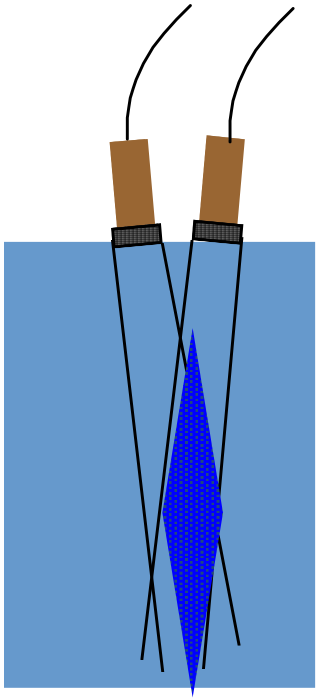
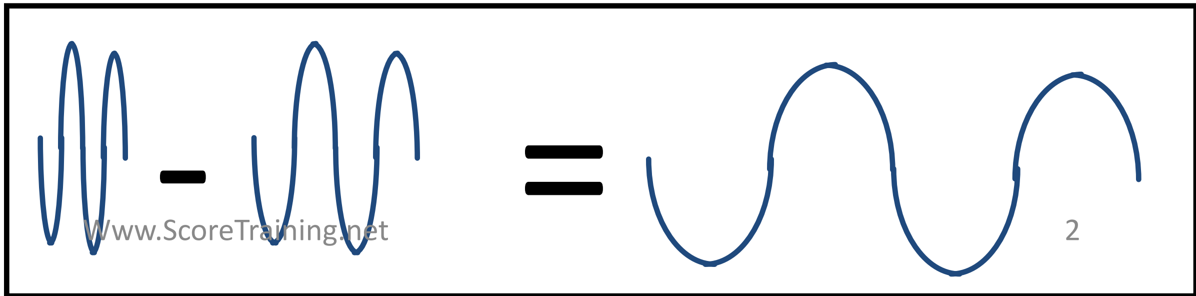
Types of doppler

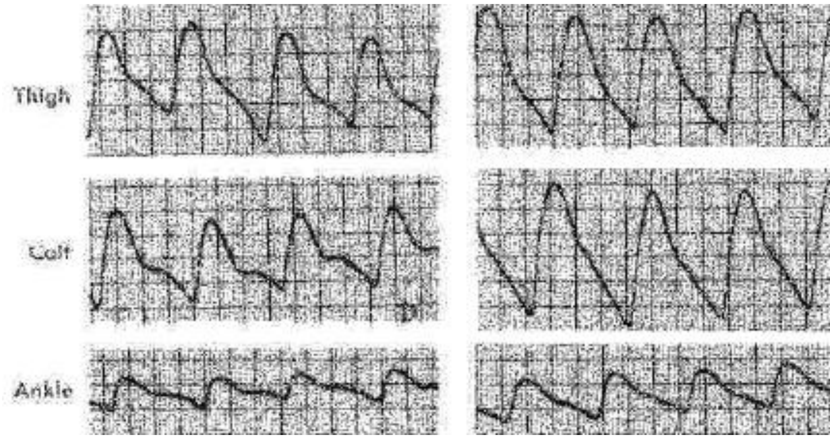
1- continuous wave doppler



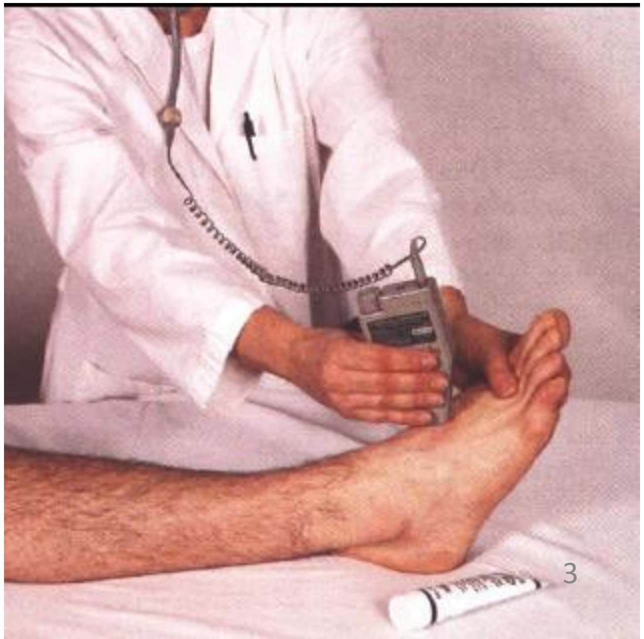
- Frequency of the transducer is chosen with operating frequency in the range of 2-10MHz (according to the vessel depth)
- The probe uses two slightly angled transducers:
  - A transmitter: energized continuously with AC with frequency F (will we use damping block?)
  - A receiver: listen continuously to the echoes coming from the sensitive volume with frequency F’
- While receiving:
  - Original frequency is suppressed
  - Doppler signal is extracted electronically from F’ received (=F-F’)



- Continuous Doppler signal can be
  - Displayed on the screen using frequency analyzer
  - Heard through loud speaker
    - ↑pitch→↑velocity
    - Harsher sound → greater flow turbulence

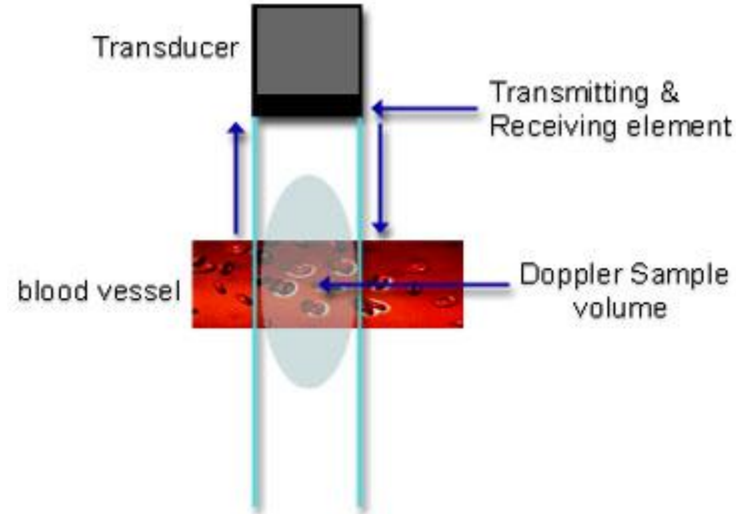


- Disadvantage:
  - Not possible to locate the moving reflector
  - Not possible to distinguish between flow of two overlapping reflectors



2- pulsed wave doppler

- Pulsed Doppler + pulsed real time B scan imaging
- Normal short B-mode pulses are interspersed with bursts of doppler pulses
- Two transducers are used:
  - Sector transducer for B-mode imaging:
    - Transmitter and receiver
    - Frame is updated once a second only
    - Short pulses are used → ↑axial resolution
    - Frequency is chosen according to the depth of the vessel
  - Pulsed doppler transducer
    - Transmitter and receiver
    - Most of the time is spent in sending doppler pulses
    - Longer pulses are used (accurate measurement of doppler shift)
    - Might operate at lower frequency



Steps of pulsed doppler imaging:

1-Real time B-mode imaging is turned on

2-a curser is set to identify the sampling volume= gated doppler acquisition area

- This is positioned over the vessel of interest
- width of sample volume = width of doppler beam

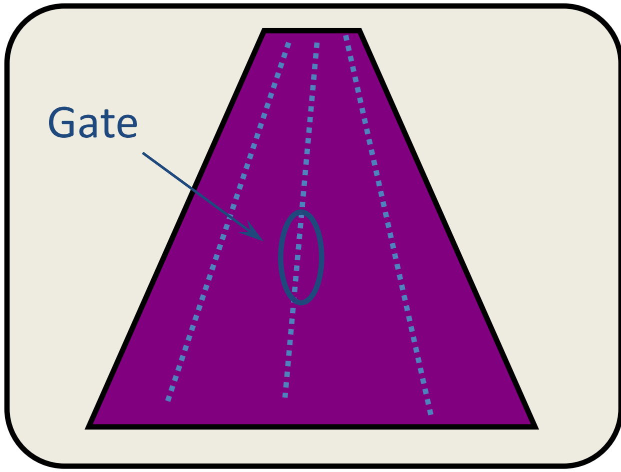
3-Range gate is set to accept only those echoes that come from the selected sample volume

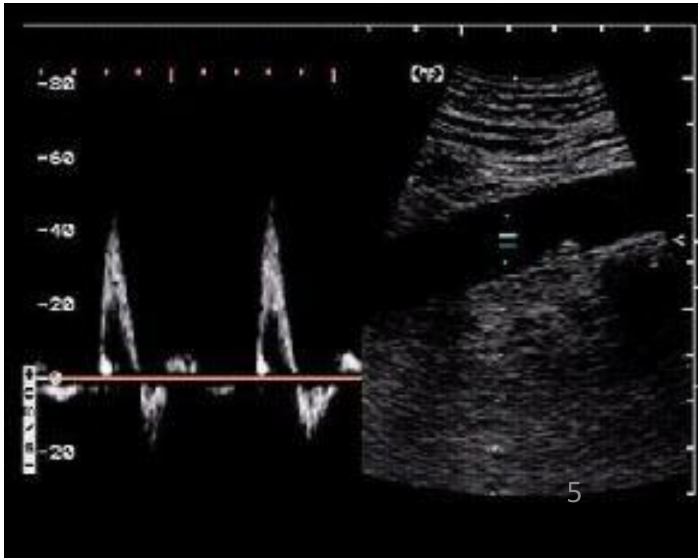
- time at which the gate opens determine the depth of tissue
- Interval for which the gate opens determines the thickness of tissues
- Example: it is known that U/S takes 7µs to travel 1 cm in tissues → if the gate opens 70 µs after the transmitted pulse , and closed after 7 µs , this means gate depth = …….. , and thickness = ……….

4-Angle θ is measured to allow calculation of blood flow velocity from the measured doppler shift

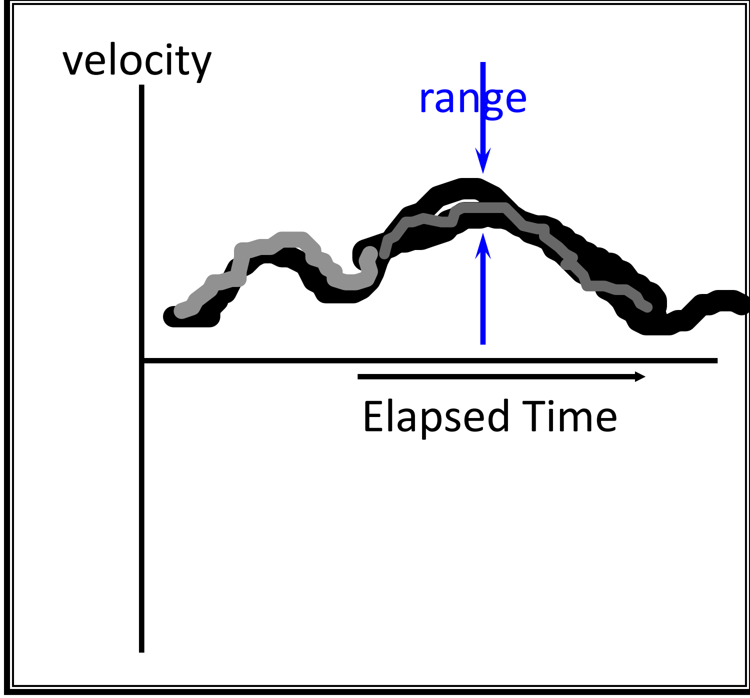
5-Interval between pulses must be long enough to image the required depth (to prevent successive doppler signals from overlapping)

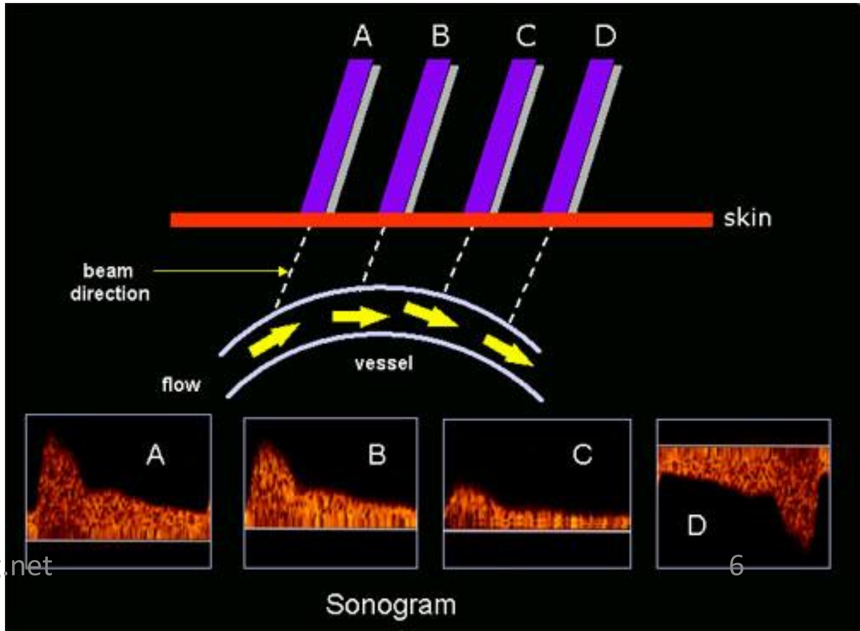
- For deep tissues: ↓PRF is used
- For superficial tissues:↑PRF is used



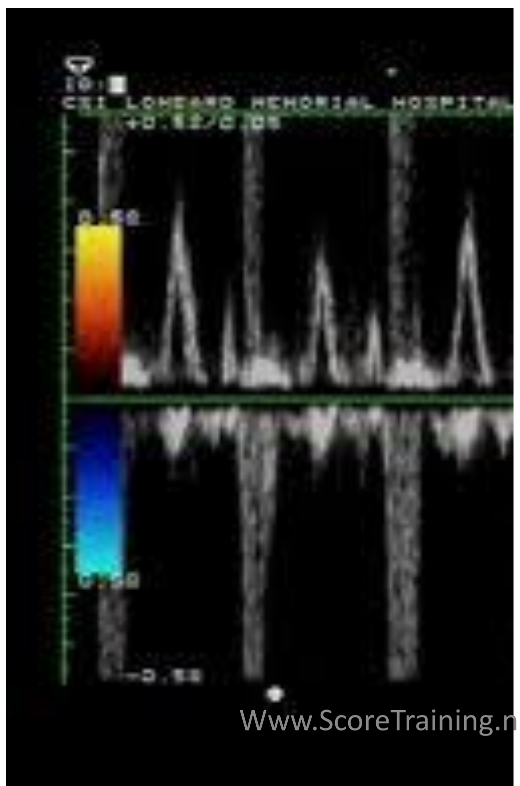


- Sonogram (spectral display) is created:
  - Definition: a graph of doppler shift (velocity) against time to determine the velocity and direction of flow at any specific time
  - It is formed of pixels each is 10 ms wide and 100 Hz tall (this means that doppler frequency is sampled and pulse is sent every 10 ms = snapshot)
  - In each snapshot , doppler signal is analyzed into its component frequencies
  - The spectrum of frequencies in each snapshot is represented as vertical column of pixels
  - Each pixel has gray level , representing number of RBCs having that velocity



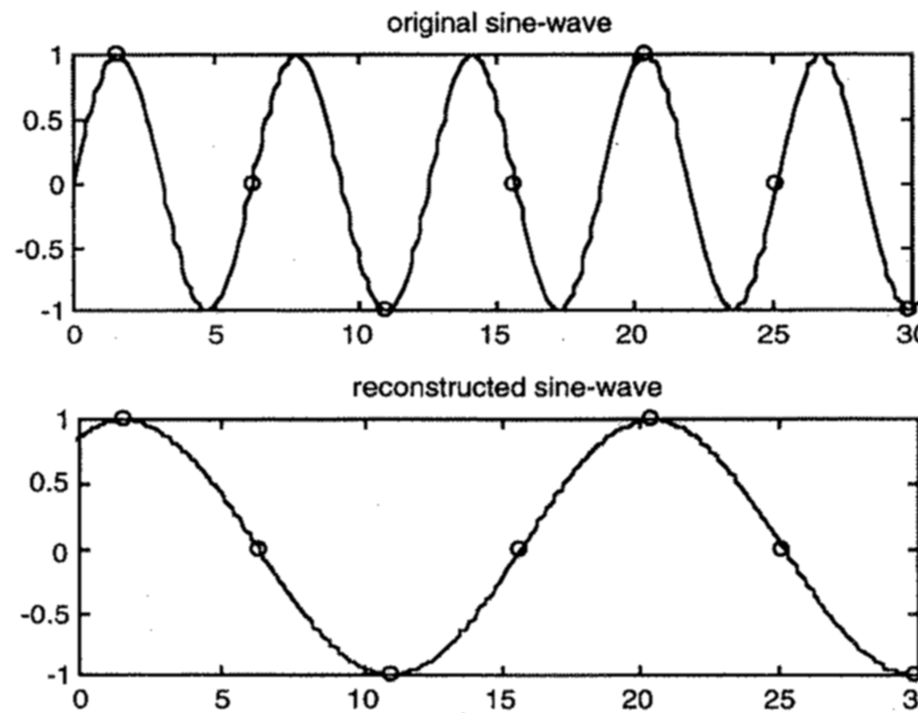


- Examples of Measurements that can be made:
  - Peak systolic velocity
  - End diastolic velocity
  - Pressure in a stensed artery:
    - bernoulli formula: $P=4V^2$
  - Volume Flow Rate = Linear flow rate X Cross Sectional Area



Nyquest law

- Any wave being measured must be sampled at least twice each cycle
- In doppler :
  - doppler pulse repetition frequency must be at least twice the maximum doppler shift frequency produced by the flow
  - i.e. the fastest flow that can be measured with accuracy is the velocity that a doppler shift frequency equal to half of the PRF being used



Pulsed doppler Aliasing

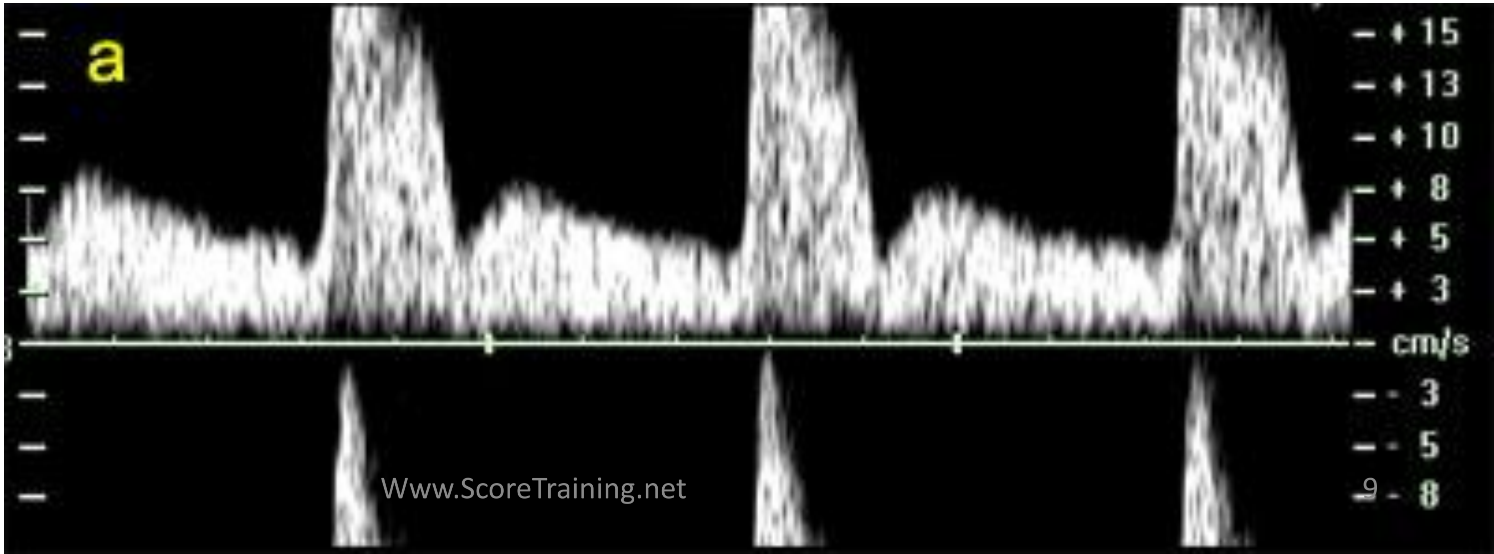
- **Definition**: an artifact which in which the flow is shown in the wrong direction and the velocity is underestimated
- **Cause**: Occur with pulsed doppler when measuring very fast flow (produce doppler shift frequency greeter than half of the PRF used)
- **Explanation**: Aliasing does not occur with continuous wave doppler
- **Methods of decreasing aliasing:**

  1- using probe of lower frequency

  2-increasing the angle θ

  Both 1& 2 will ↓ doppler shift frequency but will increased the errors of the measured flow

  3-increasing PRF



PRF , velocity , and depth

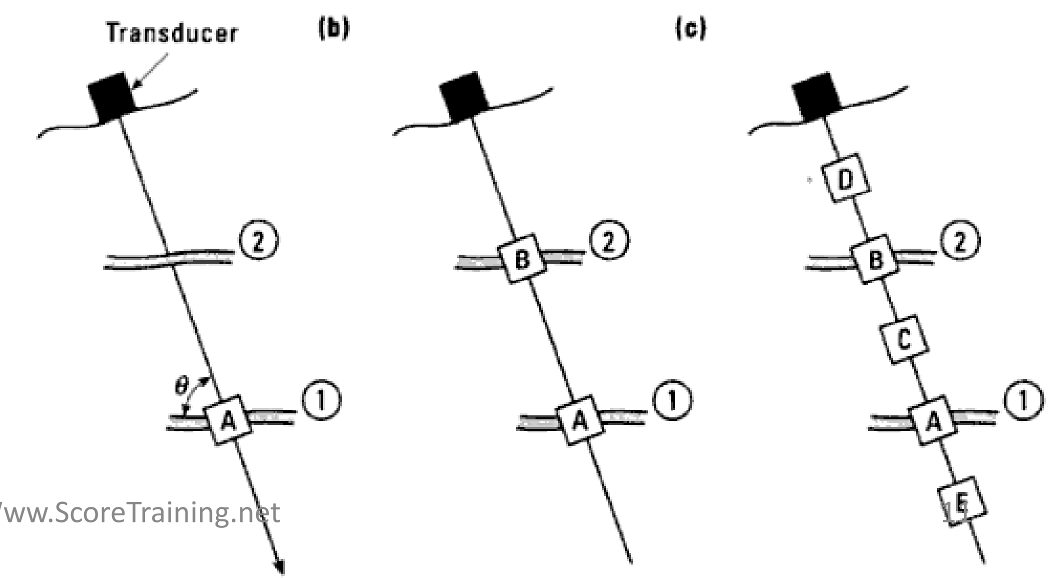
- It is difficult to measure fast flow in deep blood vessels
- Explanation
  - Deeper gate has to be used (smaller PRF)
  - So that the fastest flow that can be measured without aliasing will decrease
- Example:
  - Fast blood flow producing doppler shift of 8 KHz , must be measured by PRF of at least……….
  - This allows listening time of 60 μs, in which sound will travel through depth of ………..
- Equation:

Maximum velocity can be measured (cm/s) x range (cm) x transducer frequency(MHz)= 4000

High pulse repetition frequency mode

- Increase PRF which will cause Increase of the maximum measurable blood flow velocity
- Disadvantage:
  - decreased accuracy of the range (depth) measured (range ambiguity)
- Explanation:
  a) To measure velocity in a selected sample volume (a) : PRF is set so that there are just time to collect doppler echo from A before the transducer is pulsed again
  b) If PRF is doubled (|to increase the measurable velocity) → sonograms from A & B would superimpose
  c) If the PRF is doubled again (furtherly improve velocity measurement) → sonograms from A, B, C &D would superimpose

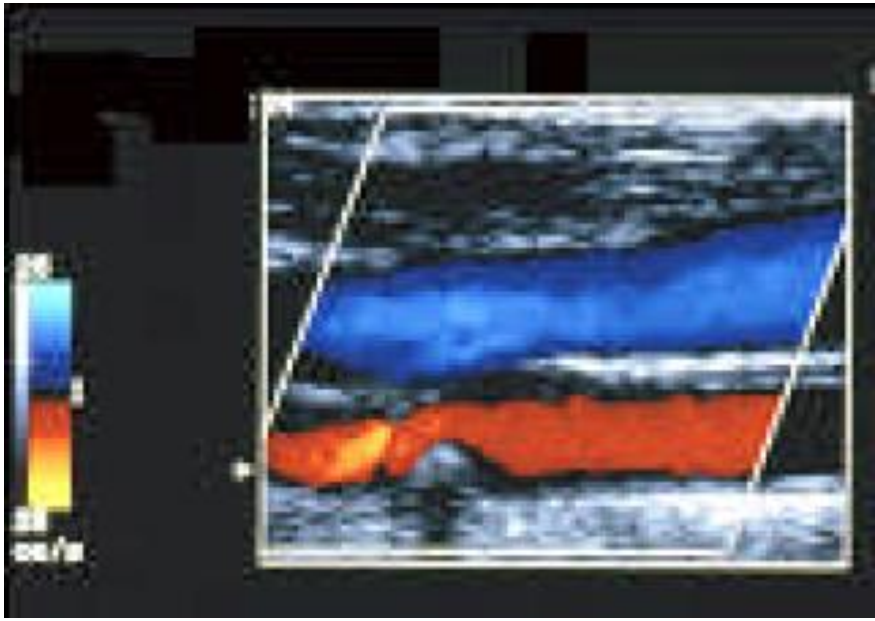
  i.e. ambiguity of the velocity is replaced by ambiguity of the range
- By the same concept : continuous wave doppler will no have aliasing , but will not have any range data



Colour flow imaging (colour mapped doppler imaging)

- **Idea**:

show direction and velocity of flow occurring in each pixel by a colour code:



A) direction:

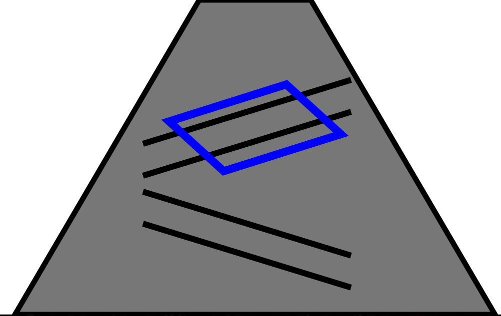
- Red : towards transducer
- Blue: away from transducer
- Yellow: turbulent

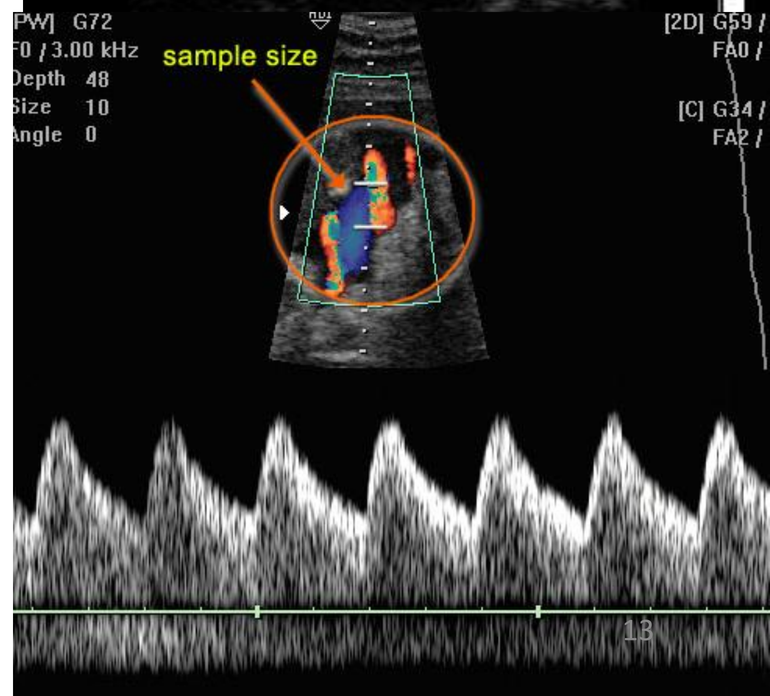
B) velocity:

- By depth of each colour

Steps of Colour flow imaging :

- Each scan line will dwell long enough for:
  - One B-scan pulse
  - followed by train of 4-12 repeated doppler pulses
- Each scan line contain 128 separate gated ranges (1.5 mm each) from which echoes are received
- Average velocity evaluated in each sample volume is be represented as colour in each pixel
- To produce sonogram: single sample volume is selected from the colour scan with switching to the spectral doppler mode





- **Disadvantage:**

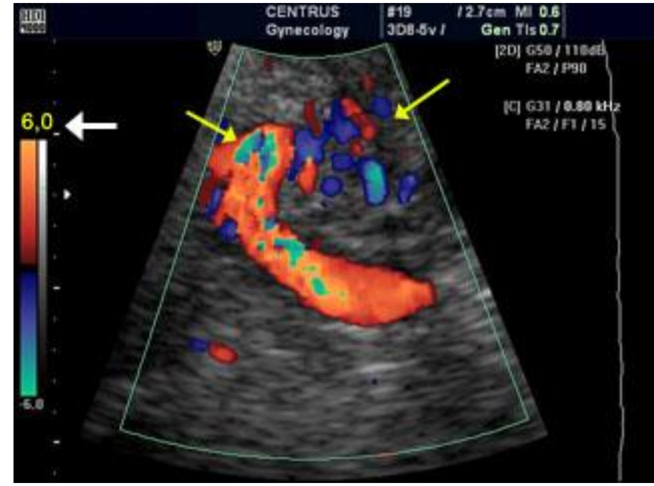
limited performance due to Limited time available to collect data:

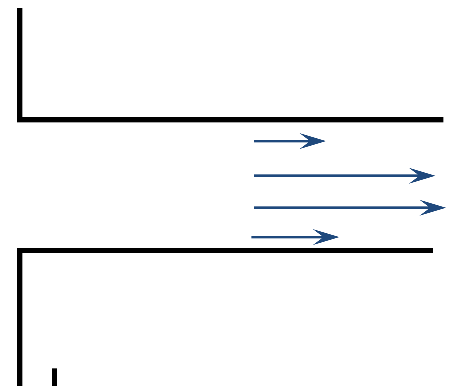
  - There are only 4-12 doppler pulses along each scan line , compared to 100 pulses in gated doppler image to produce sonogram and infinite number of pulses in continuous wave doppler
  - this is sufficient only to measure mean velocity and variance of velocity

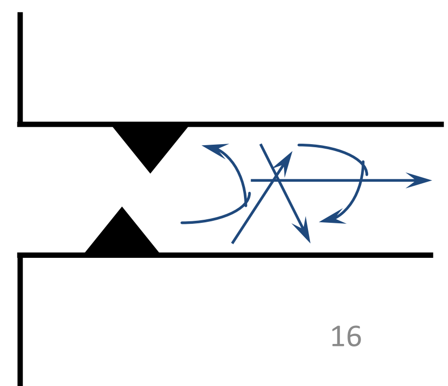
- For more accurate velocity information, some Factors that can be changed in colored doppler pulses :
  - **Frame rate** : but should be fast enough to follow changes in flow velocity
  - **Depth**: decrease it to increase PRF
  - **Field width (sector angle):** typically 30-90^{o}
  - **Line density:** should be high enough for good spatial resolution
- N.B. previous factors are interrelated , examples:
  - in children ↑frame rate is desired with↓ depth
  - We can restrict colour mapping to selected part of B-scan to achieve required frame rate and number of scan lines

Colour doppler aliasing

- If PRF is not high enough : aliasing will occur in the form of change in the expected colour of the scanned vessels
- High velocity laminar flow appears with aliased blue center , and non aliased red edges
- Turbulent jets → coloured mosaic



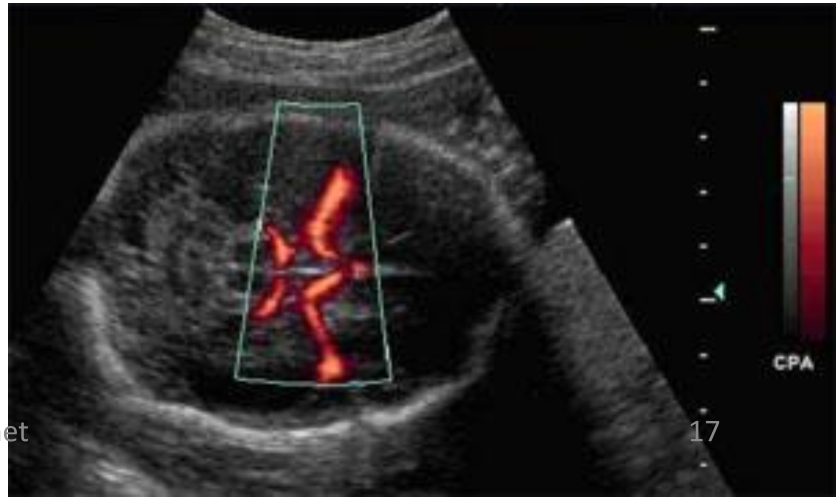




N.B. after locating the high velocity feature by colour doppler mode , machine could be switched to continuous wave doppler for accurate measurement

Power doppler

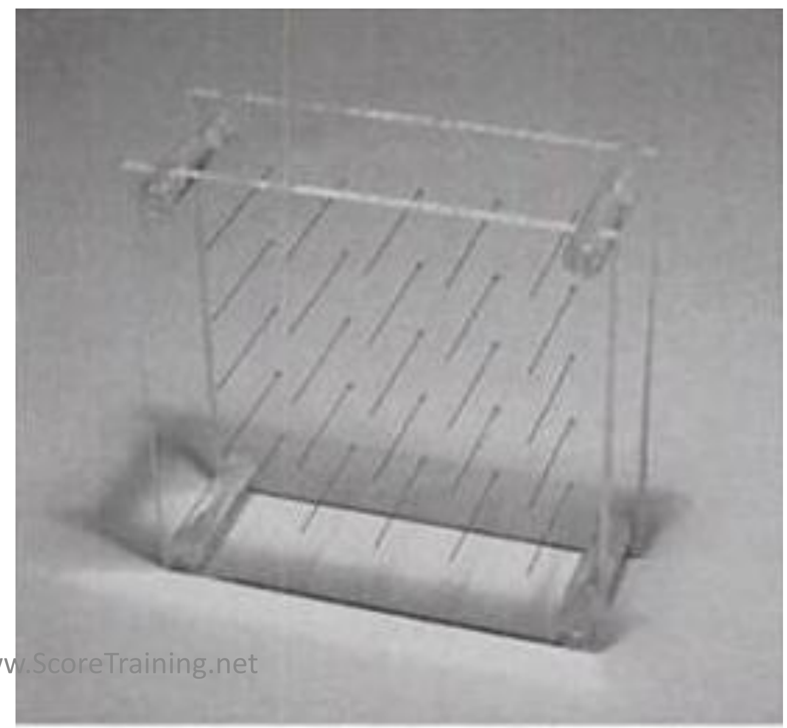
- Idea:
  - mapping amplitude of doppler signal (which depend on number of flowing RBCs) i.e. quantity of flow
- Disadvantage:
  - does not indicates the direction or velocity of flow
  - Tissue motion produce artifacts
- advantage:
  - high sensitivity to small vessels and stagnant flow →differentiate between areas of flow and areas with no flow
  - Less dependant on insonation angle
  - Not subject for aliasing (non directional)



Ultrasound quality assurance

resolution

- Tested by a rig composed of parallel wires mounted on a frame and immersed in Perspex path containing a fluid (in which sound velocity is the same as in tissues)



Sensitivity , dynamic range and accuracy

- Perspex block with equally spaced vertical rods of different diameters
- Each 7 mm of Perspex is equivalent of 4 mm of tissues (sound is faster in Perspex)

Power output

Two methods

1- weighing the sound with force balance

2- measuring heating effects using a calorimeter

Gray scale performance and doppler function

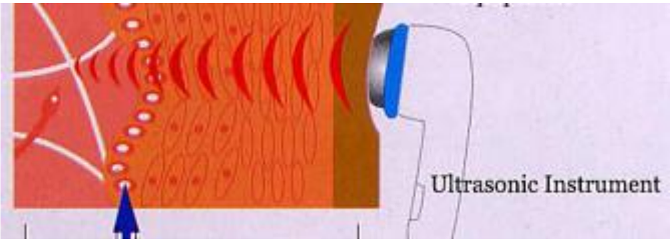
- Tissue equivalent phantoms based on gelatin is used

Ultrasound safety

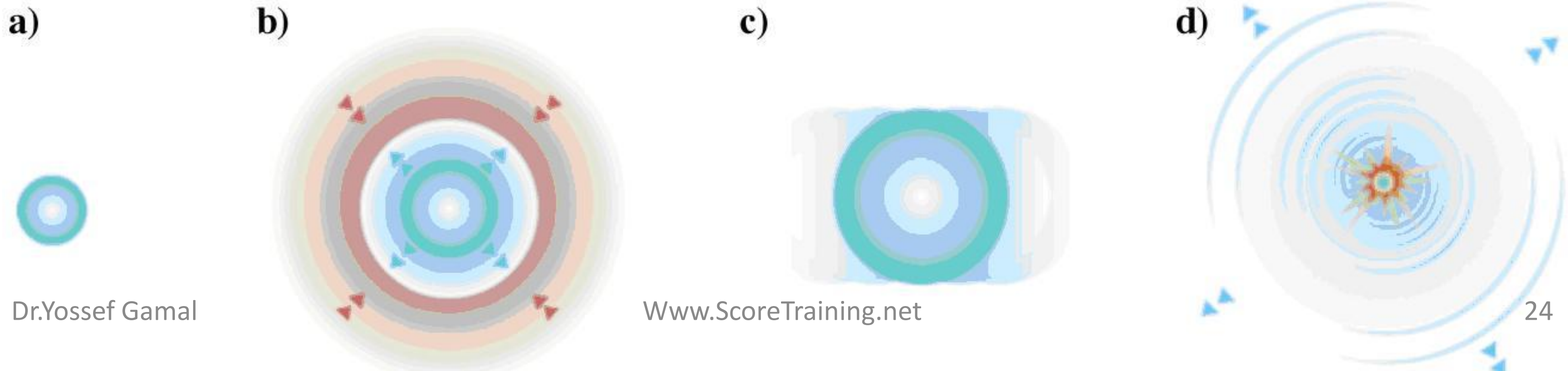
- Ultrasound is non ionizing and of low risk
- Harmful effects are more linked to ultrasound therapy (no confirmed evidence of damage from diagnostic ultrasound exposure)

Ultrasound risks

- **Local heating** due to frictional , viscous and molecular relaxation processes leading to chemical damage (but also is used for therapy)
- **Acoustic streaming** of cellular contents in the direction of the beam affecting cell membrane permeability



- **Violent acceleration of particles** → mechanical damage of cell membranes
- **Cavitation**:
  - the pressure changes cause micro-bubbles in a liquid to expand this may lead to sudden collapse → enormous rise in temperature ,
  - this is more likely to occur at high pressures and low frequencies , and less likely to occur in pulsed beam (each pulse will not last long enough for resonance to be reached)



Safety guidelines

- Time average intensity should not exceed $100 mW/cm^2$
- Total sound energy (intensity x time) , should not exceed $50\ J/cm^2$

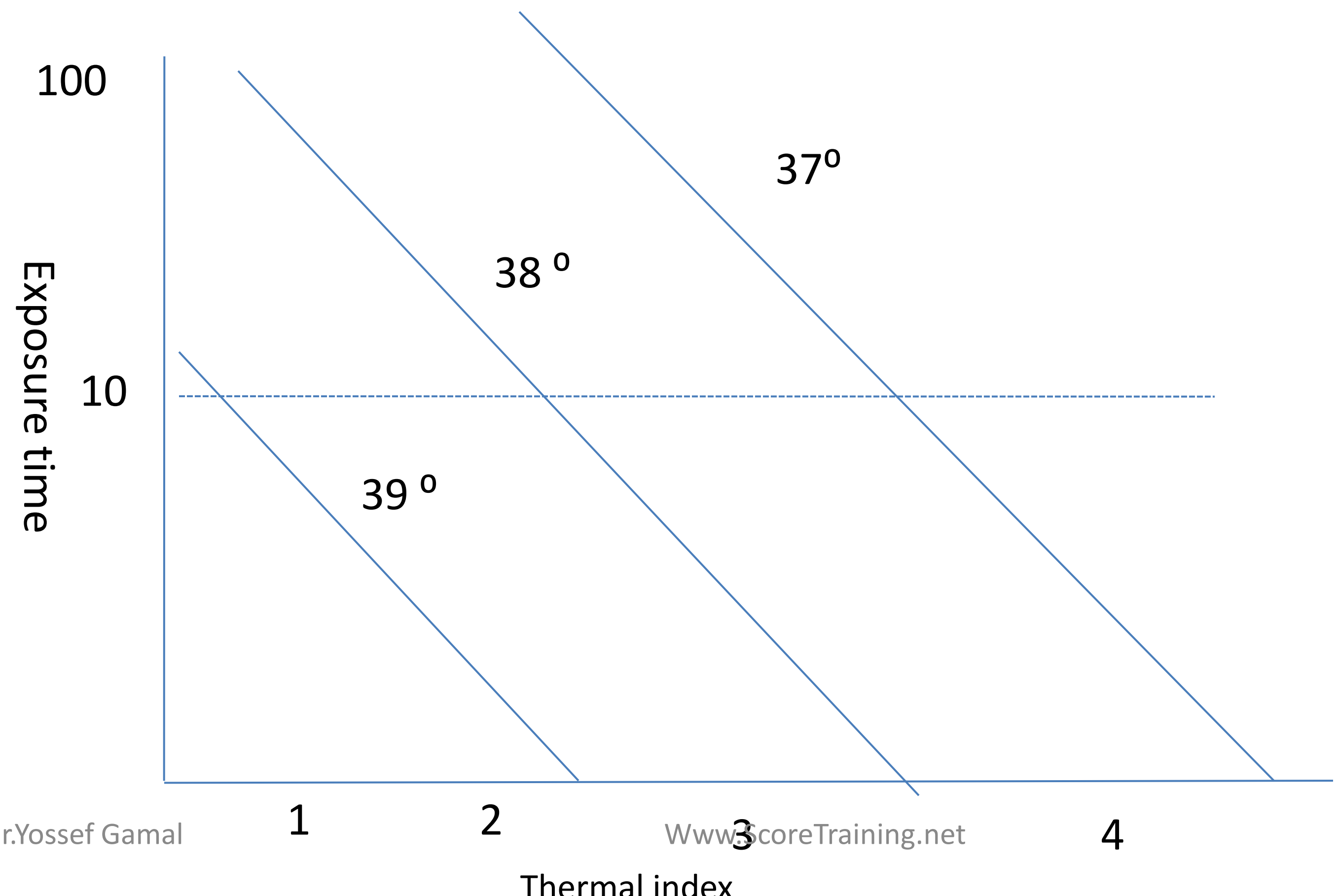
Thermal index

- Ratio of the power emitted to that required to increase the temperature by 1^{o}C
- Gives indication to temperature rise in tissues

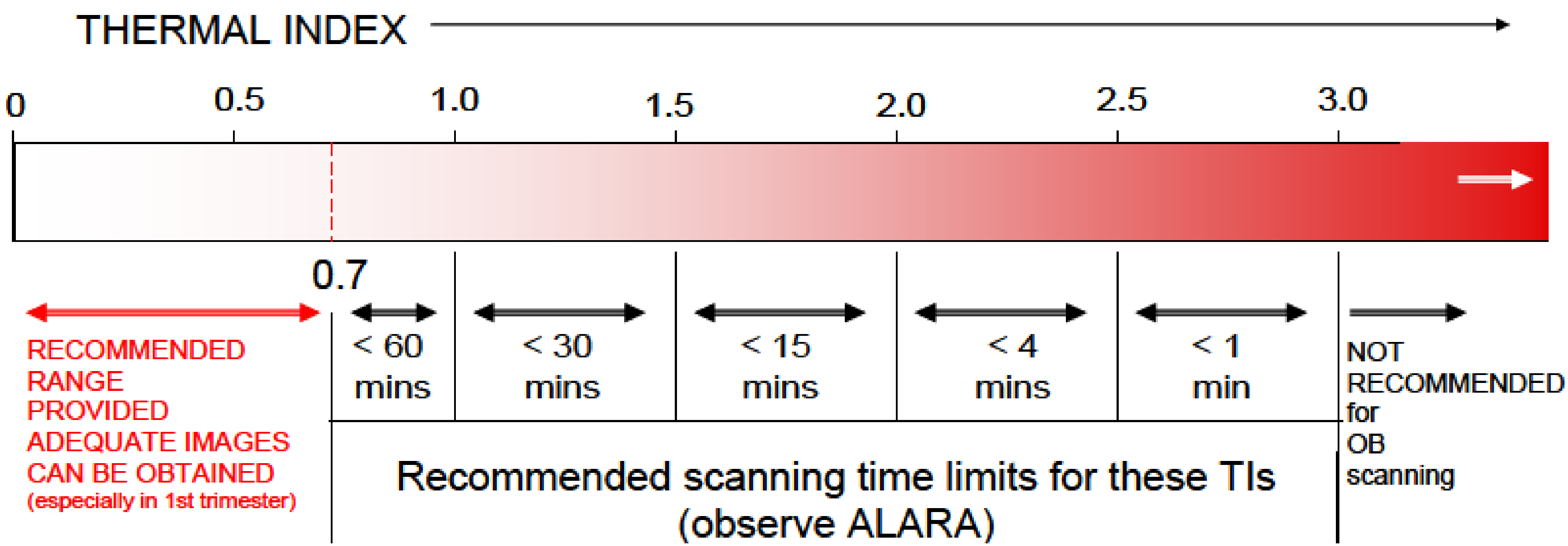
Mechanical index

$$= \frac{\text{Peak of rarefaction pressure}}{F^2}$$

- Both indices are indicated on the scanner display panel when the equipment itself can exceed index value of 1
- Thermal Index values less than 0.5 are below the threshold level of any unsafe effects
- Exposure times should reduced especially when scanning feverish patients and when using doppler systems



Obstetric ultrasound safe limits



Basics of Magnetic resonance imaging

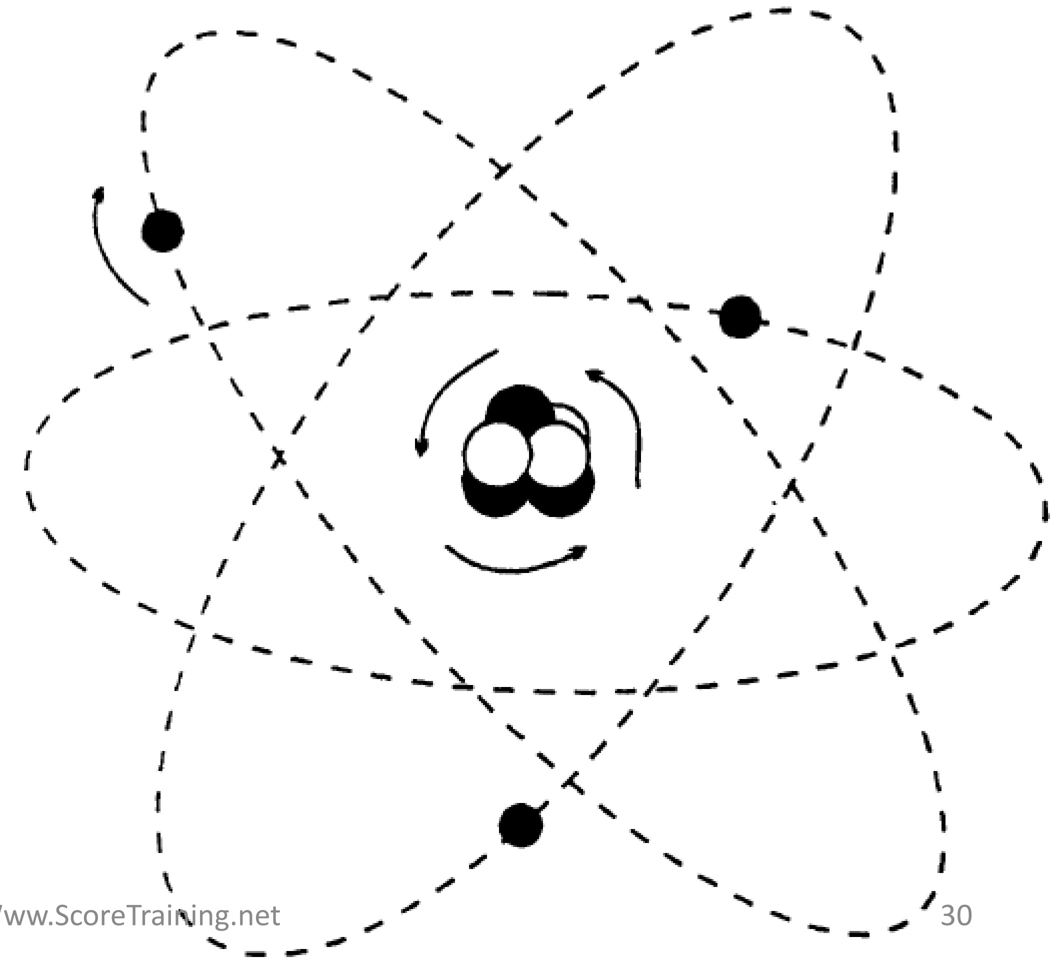
- Three types of motion are present within the atom . These are:

(1) electrons spinning on their own axis,

(2) electrons orbiting the nucleus,

(3) the nucleus itself spinning about its own axis.

- The principles of MRI rely on the spinning motion of specific nuclei present in biological tissues, known as *MR active nuclei.*



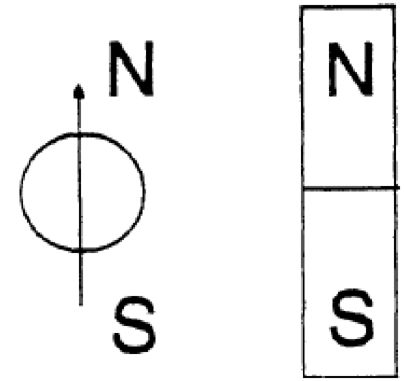
MR active nuclei:

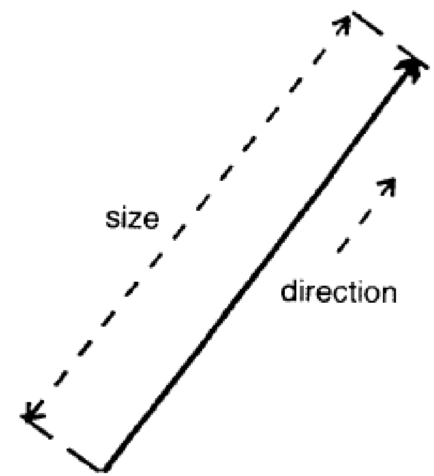
- Have tendency to align their axis of rotation to an applied magnetic field
- This occurs if the mass number is odd, i.e. even number of neutrons & odd number of protons or vice versa.
- examples of MR active nuclei:
  - – hydrogen 1
  - – carbon 13
  - – nitrogen 15
  - – oxygen 17
  - – fluorine 19
  - – sodium 23
  - – phosphorus 31

The hydrogen nucleus

- The hydrogen nucleus is the MR active nucleus used in clinical MRI.
- contains a single proton (atomic and mass number 1).
- It is used because
  - it is very abundant in the human body,
  - Its solitary proton gives it a relatively large magnetic moment.

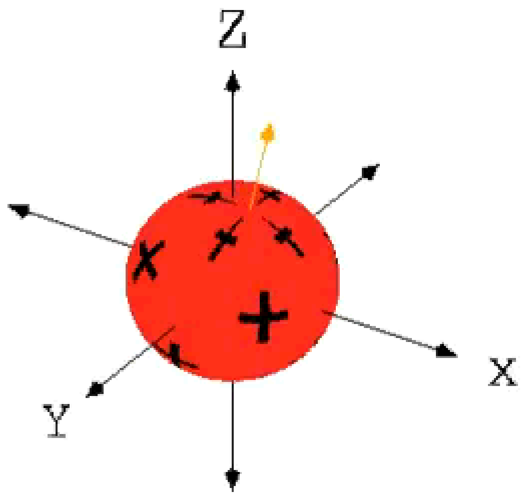
- The hydrogen nucleus as a magnet
  - a magnetic field is created when a charged particle moves. Therefore the hydrogen nucleus (a spinning proton) has a magnetic field and acts as a small magnet (with a north & a south poles)= magnetic moment
  - The magnetic moment is denoted by an vector (direction = direction of the magnetic moment, length= size of the magnetic moment)

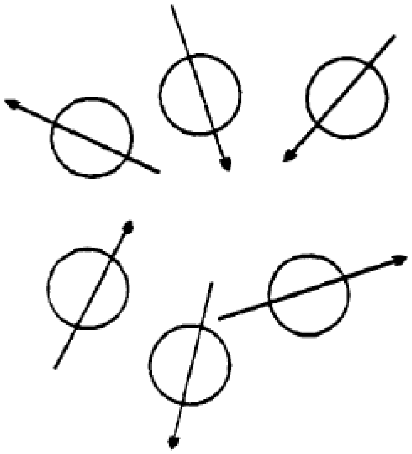




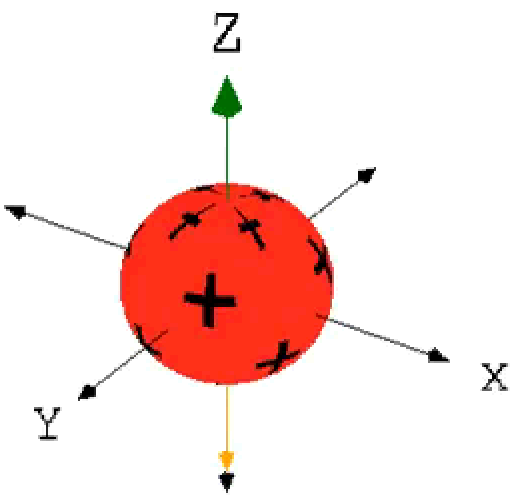
Alignment of hydrogen protons

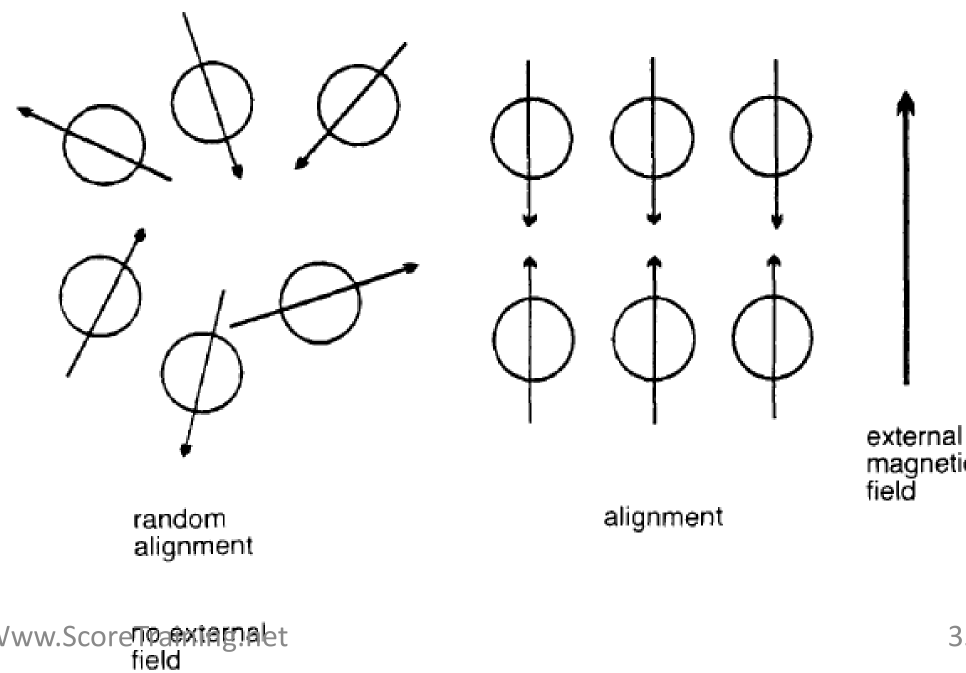
- In the absence of an applied magnetic field→
  - the magnetic moments of the hydrogen nuclei are randomly orientated.





- When placed in a strong static external magnetic field B_0 →
  - the magnetic moments align with this magnetic field.
  - Some protons align parallel with the magnetic field ,while a smaller number of the nuclei align anti-parallel to the magnetic field (in the opposite direction)



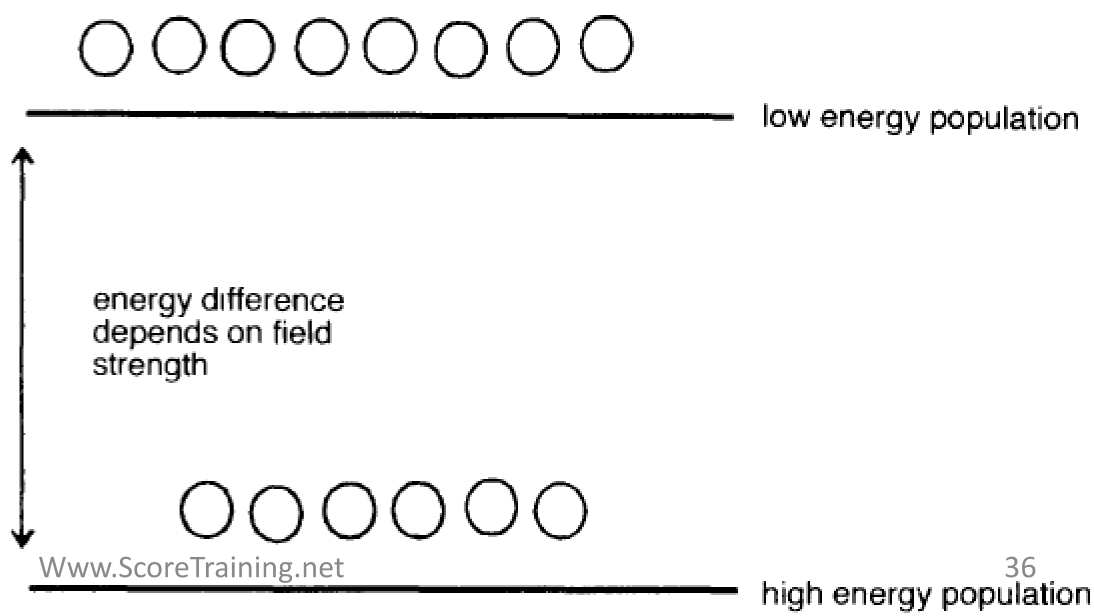


i.e. hydrogen nuclei only possess two energy states:

- Low energy nuclei: align their magnetic moments parallel to B_0 (spin up nuclei)
- High energy nuclei: align their magnetic moments in the anti-parallel direction (spin down nuclei).

Energy difference between them is determined by the field strength:

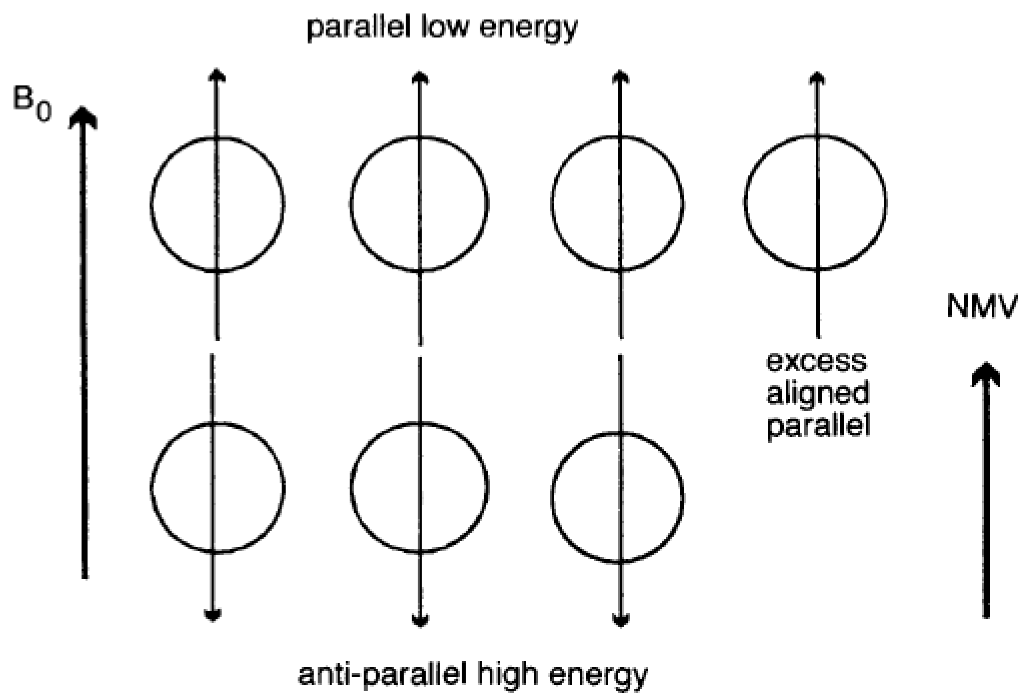
- As the strength of B_0 field increases, fewer nuclei have enough energy to oppose it



- there are always fewer high energy nuclei → the magnetic moments of the parallel nuclei cancel out the anti-parallel moments

i.e. there is always a small excess in the parallel direction producing a net magnetic vector in the same direction

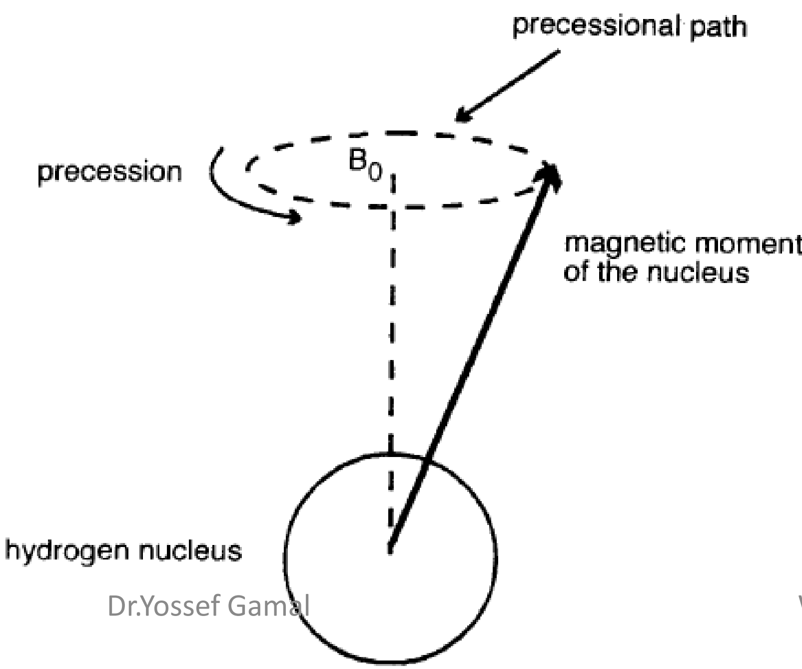
- When a patient is placed in the bore of the magnet, the hydrogen nuclei within the patient align parallel and anti-parallel to Bo. With The energy difference between the two populations increases as Bo increases.
- i.e. the magnitude of the net magnetization vector (NMV)is larger at high field strengths resulting in improved signal.

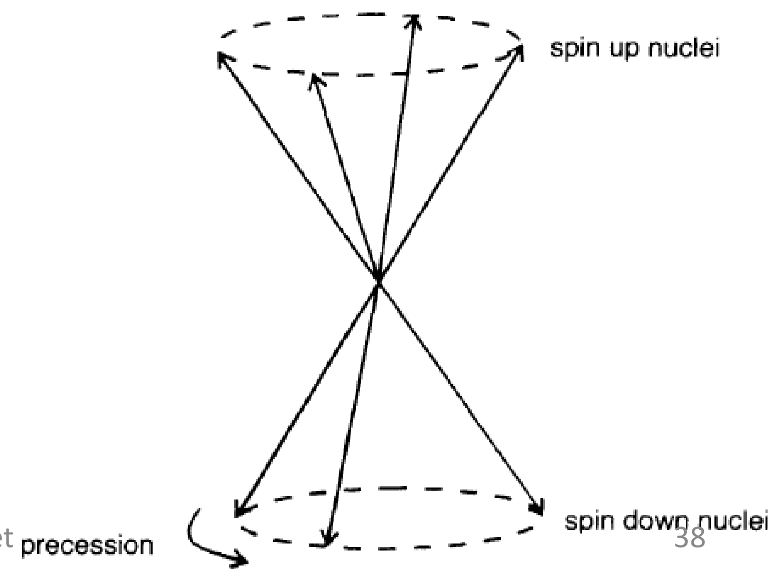


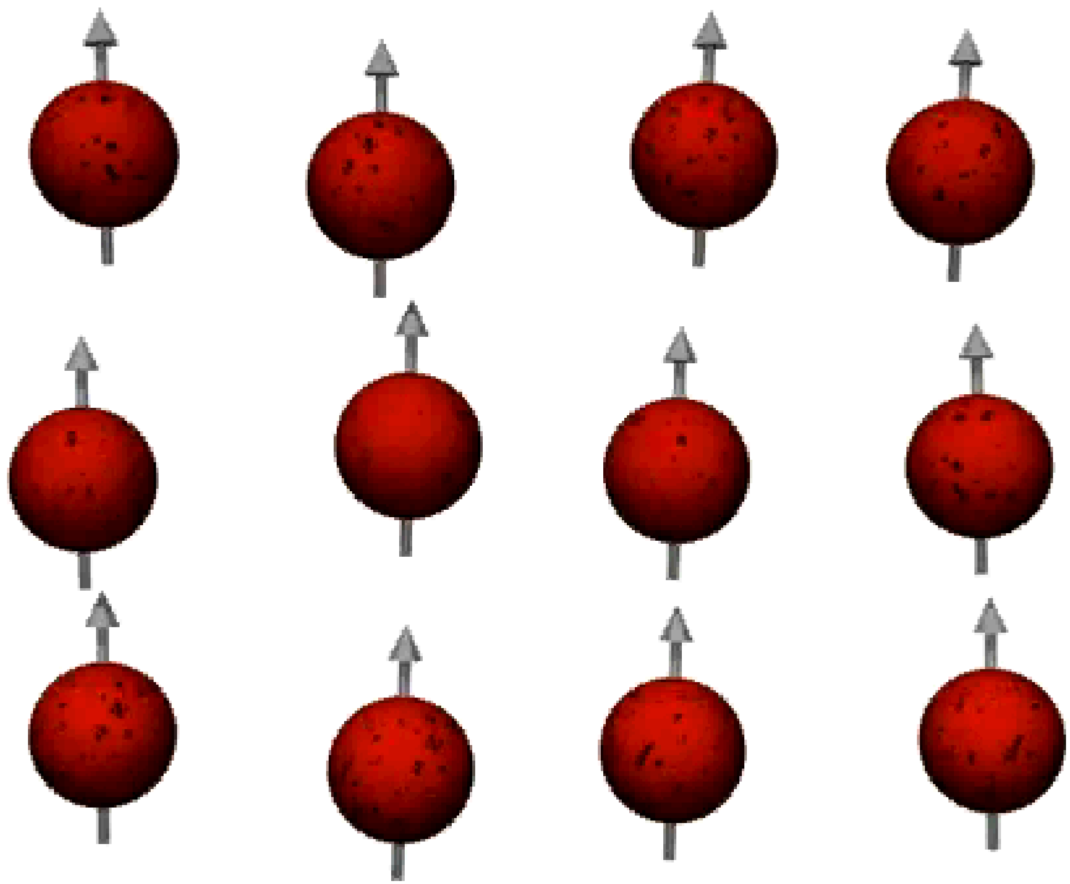
N.B. The unit of Bo is tesla (T) or gauss (G). (1 tesla = 10000 gauss)

Spinning and Precession

- Types of H nucleus movement when Bo is applied:
  - **spinning** around its axis (the normal movement)
  - **Precession** of the NMV around Bo
    - only occur when Bo is applied
    - the speed at which the NMV wobbles around Bo is called the precessional frequency.
    - The unit of precessional frequency is megahertz (MHz)







The Larmor equation
the precessional frequency (ω_0)= B_0 x γ

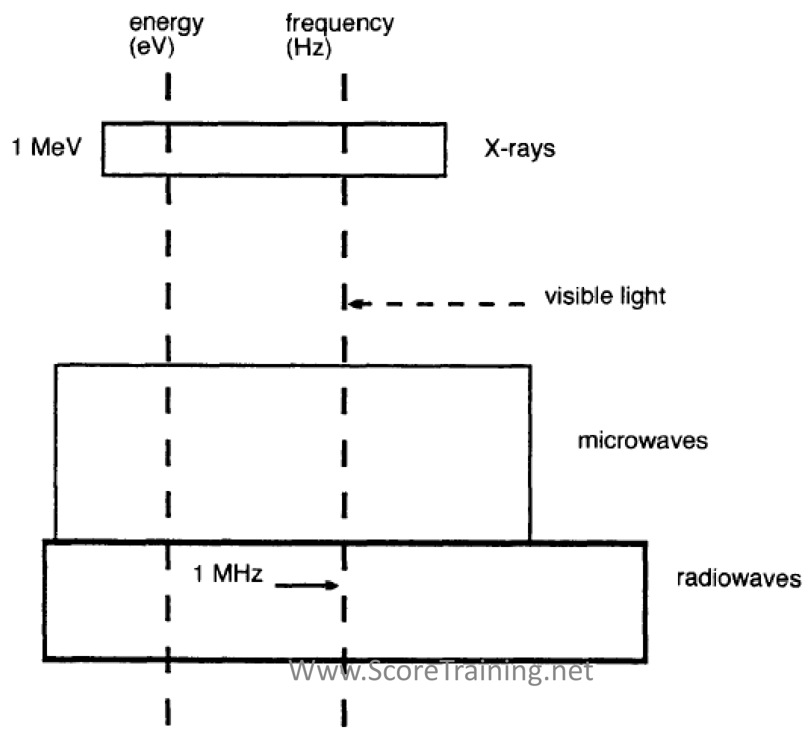
- γ= gyro-magnetic ratio.
  - It is constant for each MR active nucleus (=precessional frequency at 1.aT)
  - The gyro-magnetic ratio of hydrogen is 42.57 MHz/T.
  - This means that MR active nuclei have different precessional frequencies at the same field strength.
- The precessional frequency is often called the *Larmor frequency*
- Larmor frequency is proportional to the **Bo** .
- hydrogen has a different precessional frequency at different field strengths. For example:
  - at 1.5 T the precessional frequency of hydrogen is 63.86 MHz(42.57 MHz x 1.5 T)

Resonance

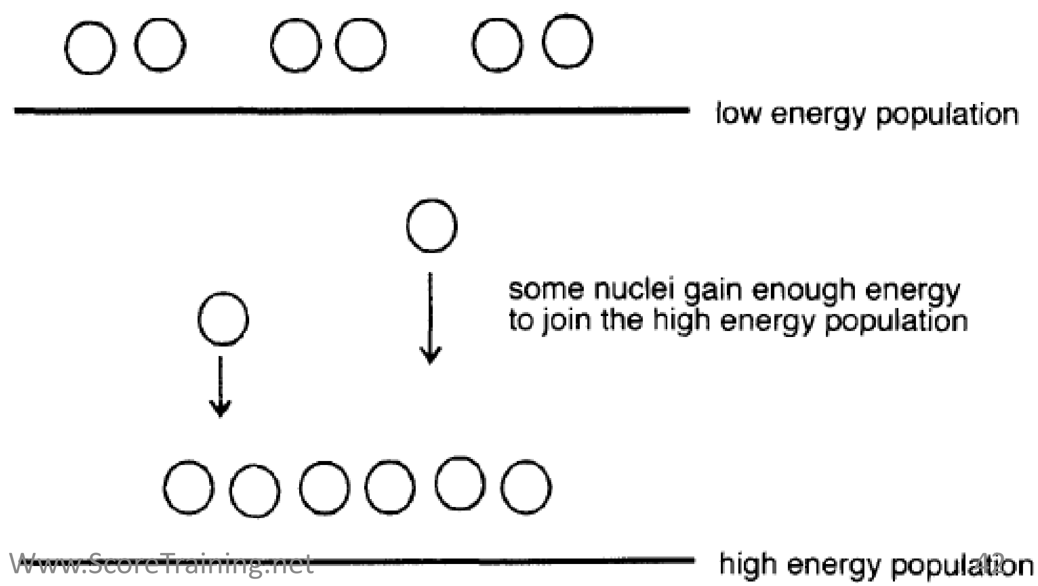
- When a nucleus is exposed to an external electromagnetic energy that has a frequency similar to its own precessional frequency, the nucleus gains energy from the external force.
- Energy at the precessional frequency of hydrogen corresponds to the radio frequency (RF) band of the electromagnetic spectrum

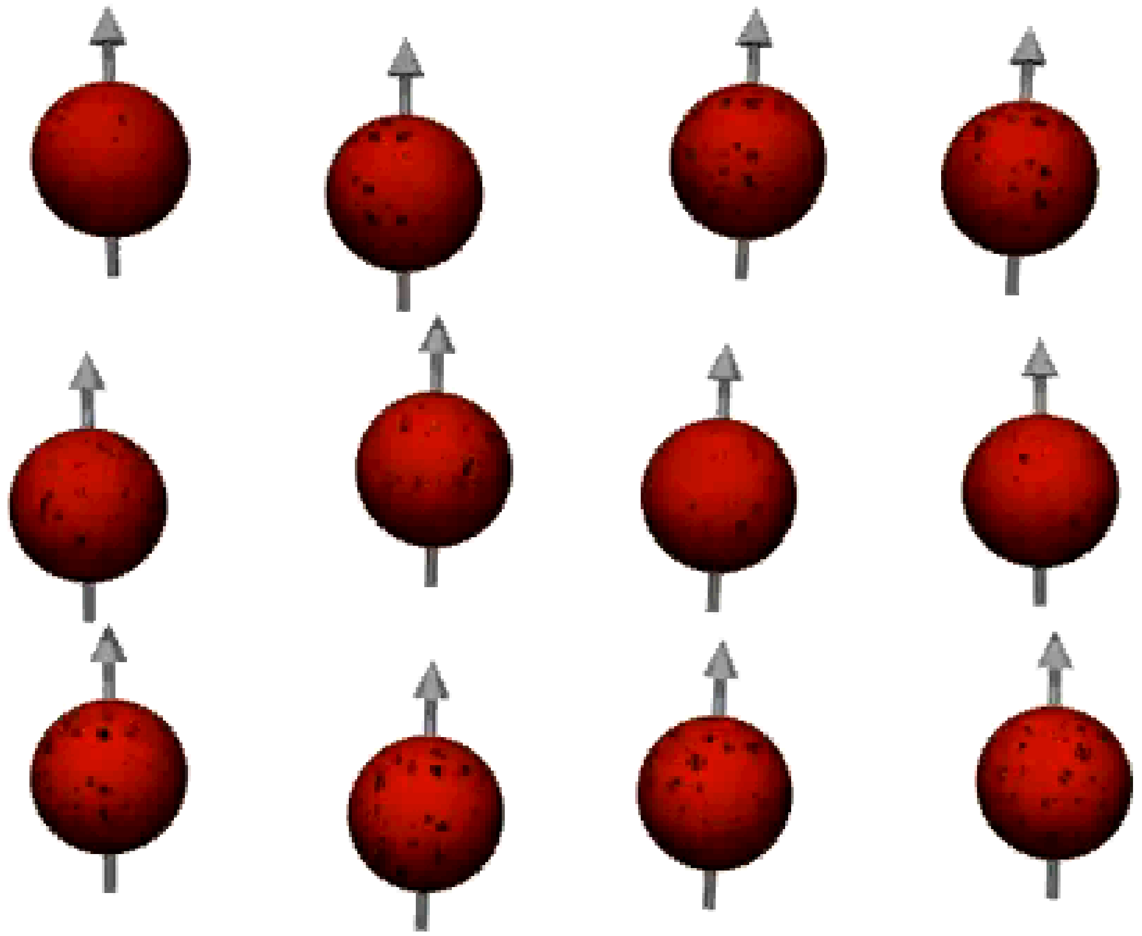
i.e. an RF pulse of energy at exactly the Larmor frequency of the hydrogen NMV, is applied.

- Other MR active nuclei that have aligned with Bo do not resonate (why?).
- The application of an RF pulse that causes resonance is termed excitation.



- This absorption of energy causes an increase in the number of spin down hydrogen nuclei populations as some of the spin up nuclei gain energy
- As the field strength increases, the energy difference between the two populations also increases so that more energy (higher frequencies) are required to produce resonance.

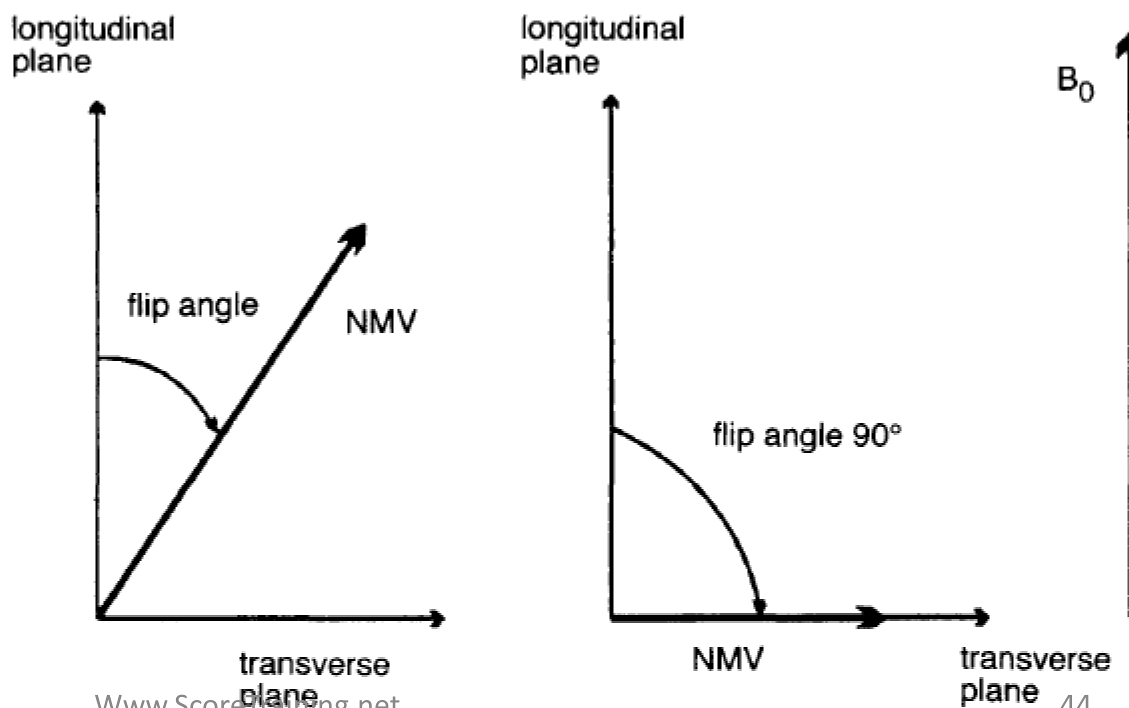




The results of resonance

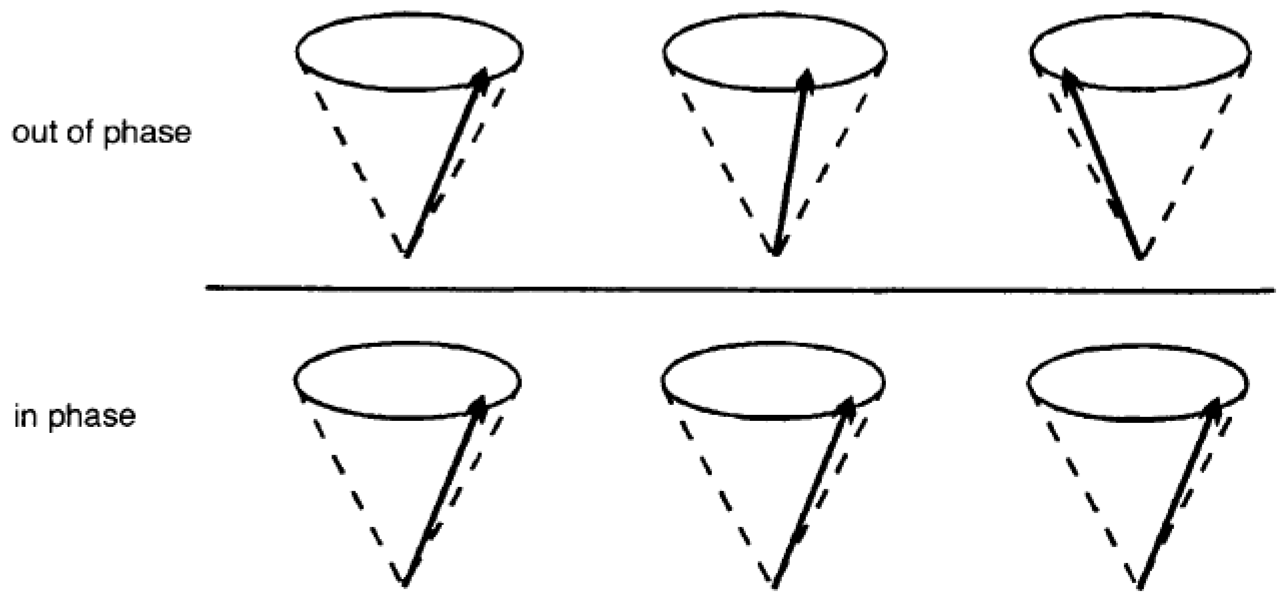
1- NMV moves out of alignment away from Bo:

- The angle to which the NMV moves out of alignment is called the *flip angle*
- *The magnitude of the flip angle depends upon* the amplitude and duration of the RF pulse.
- Bo is now termed the *longitudinal axis/plane.*
- The plane at 90° to Bo is termed the *transverse plane.*
- e.g. flip angle is 90°: nuclei is given enough energy so that the longitudinal NMV is completely transferred into a transverse NMV (rotates in the transverse plane at the Larmor frequency)



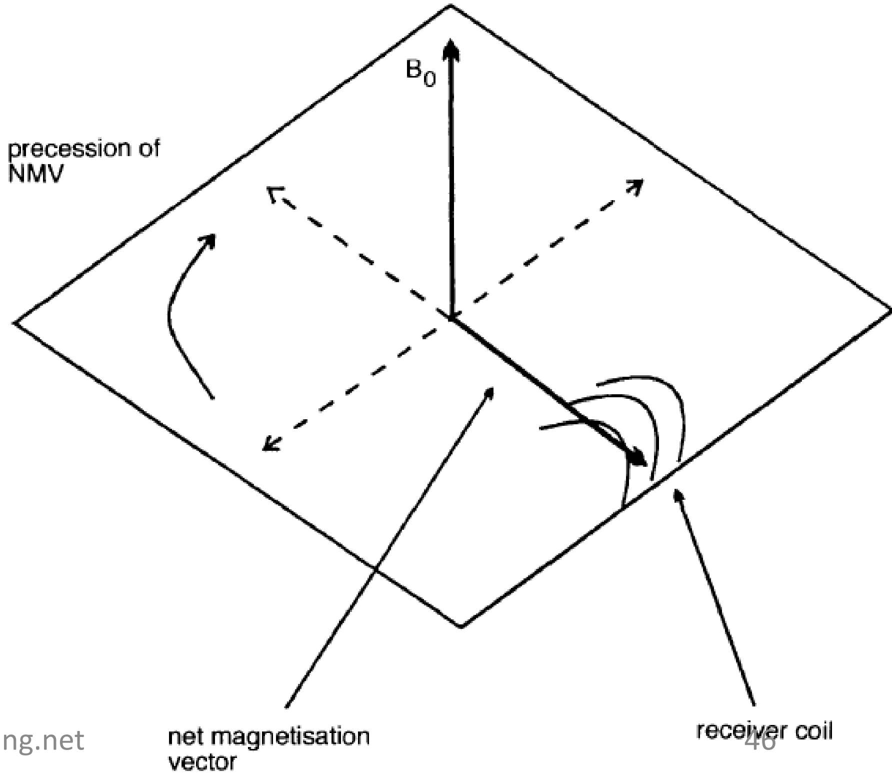
2- the magnetic moments of the hydrogen nuclei move into phase with each

– i.e. Magnetic moments *become in the same* place on the precessional path around Bo at any given time



The MR signal

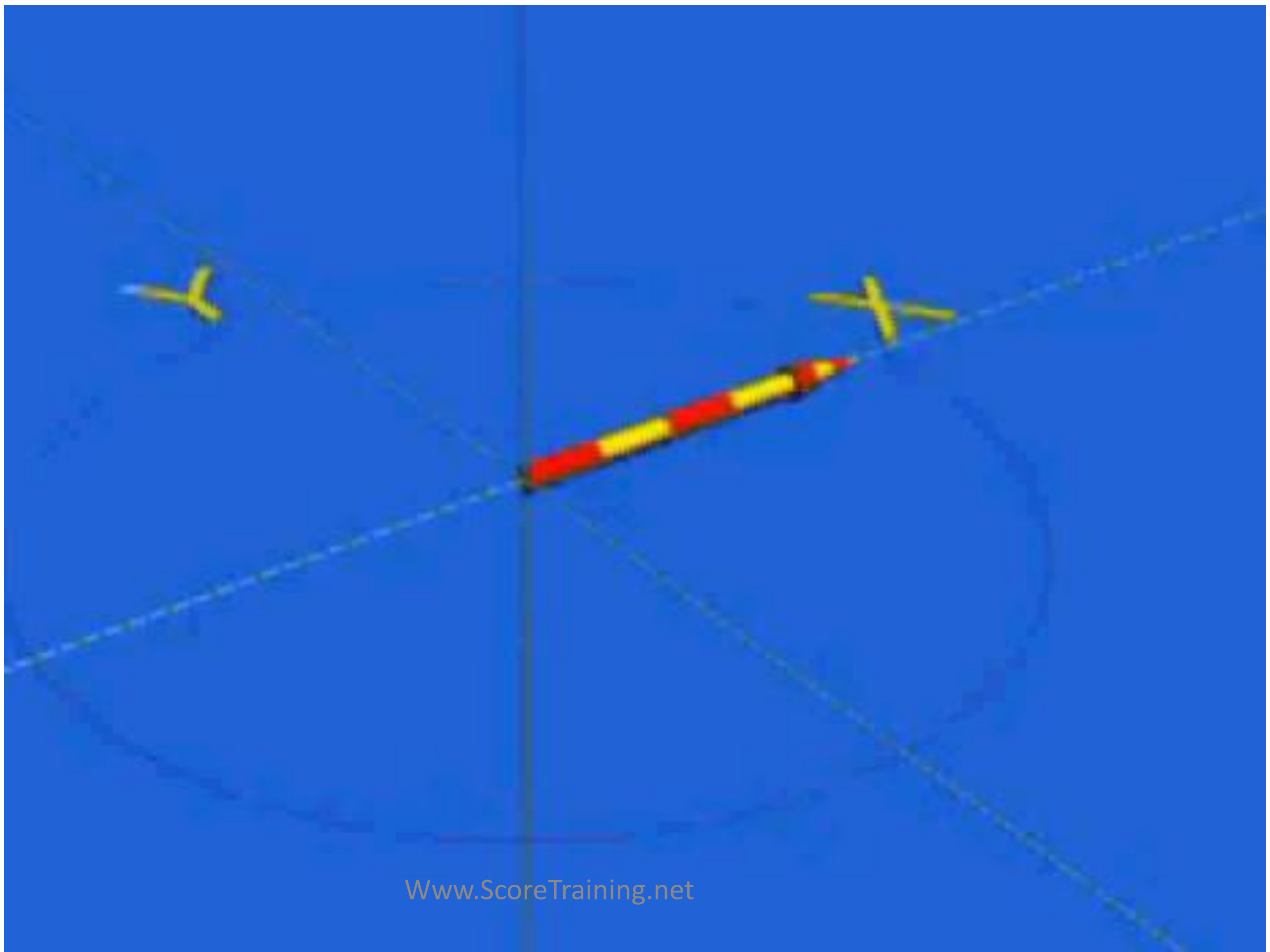
- As a result of resonance, the NMV is precessing in phase in the transverse plane.
- Faraday's laws of induction state that if a receiver coil is placed in the area of a moving magnetic field a voltage is induced in this receiver coil.
- A receiving coil is put in the transverse plane and a voltage is produced in the coil when the in-phase NMV precesses in the transverse plane
- This coil voltage = *MR signal.*
  - *The frequency of the signal is the same* as the Larmor frequency
  - the magnitude of the signal depends on the amount of magnetization present in the transverse plane.
  - MR signal is alternating



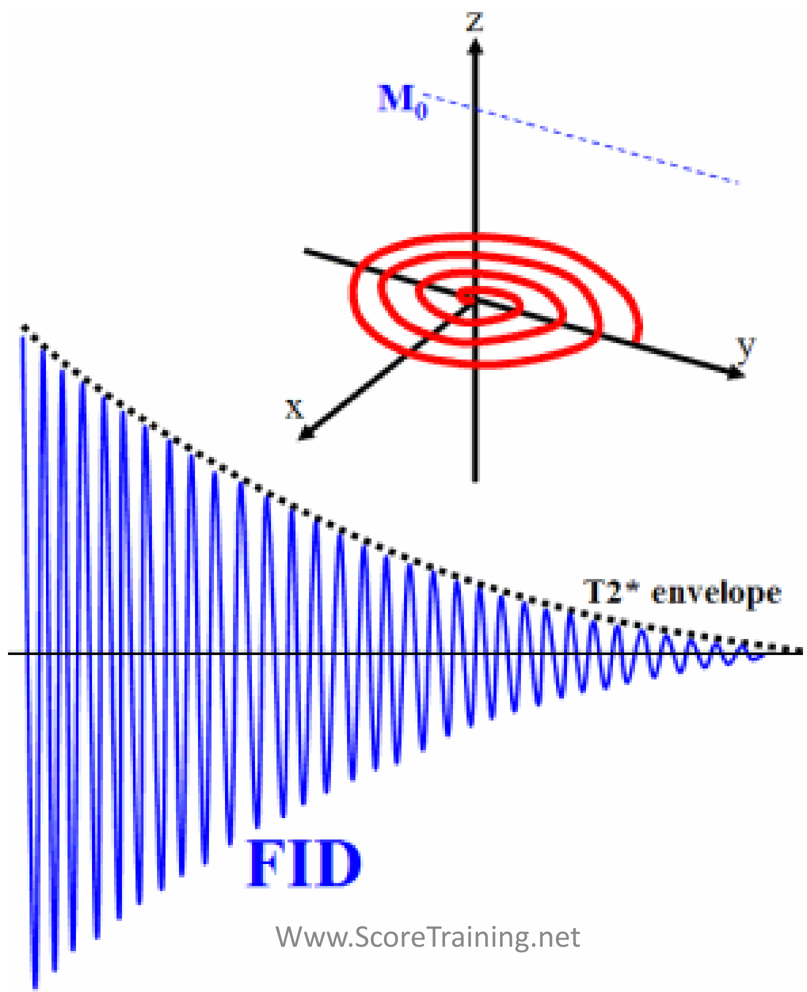
- When the RF pulse is switched off, two processes occur at the same time but independently:
  - *1)Recovery:*
    - The amount of magnetization in the longitudinal plane gradually increases (the NMV is again influenced by Bo)
    - *This occur due to **relaxation** (i.e. loss of the absorbed energy)*

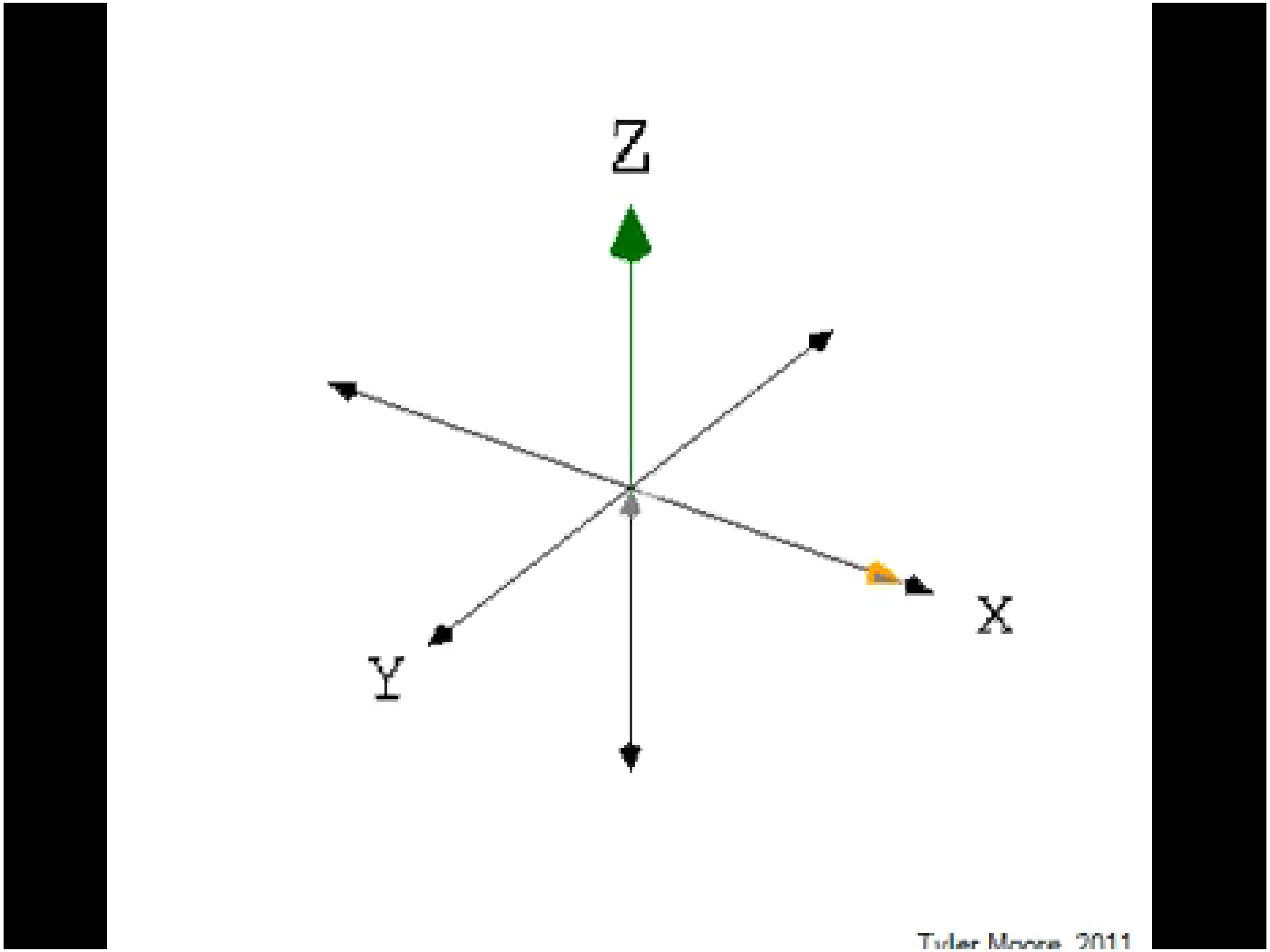
– *2)Decay*

- The amount of magnetization in the transverse plane gradually decreases
- *This occur due to* dephasing.



- As the magnitude of transverse magnetization decreases, so does the magnitude of the voltage induced in the receiver coil. The reduced signal is called the *free induction decay (FID) signal.*

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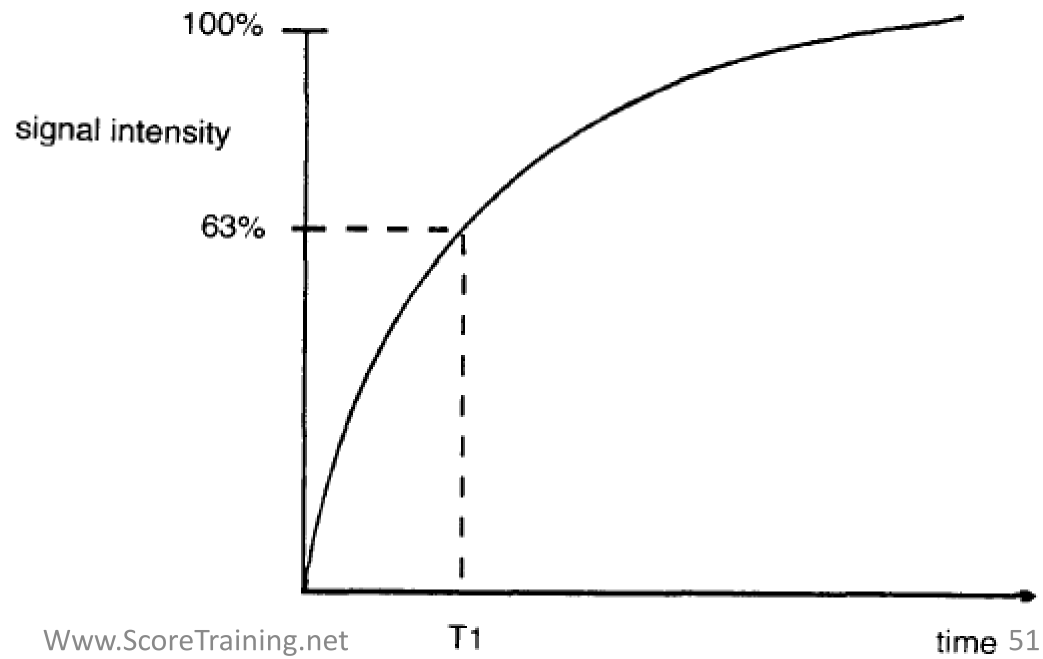
Tyler Moore, 2011

T1 recovery

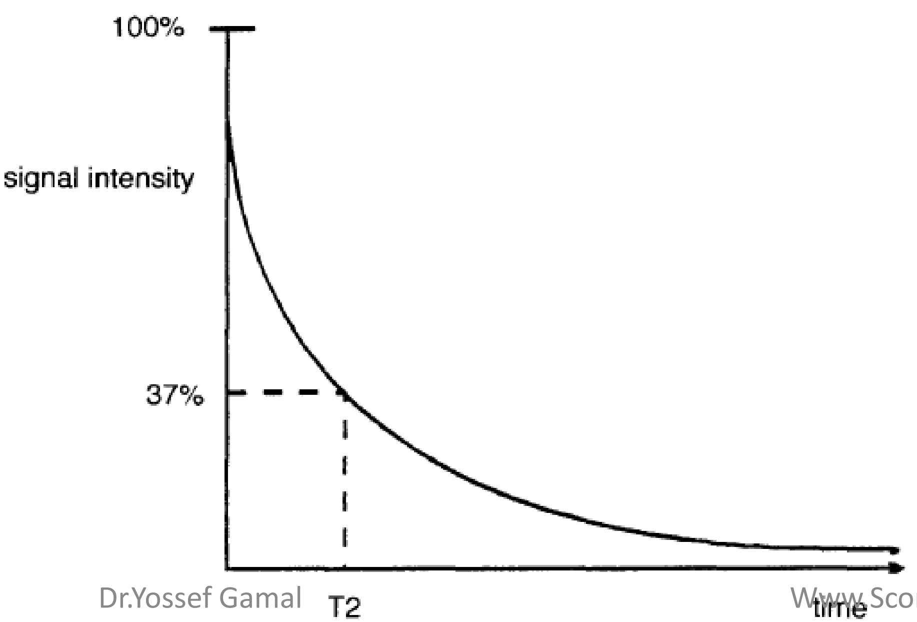
- The recovery of longitudinal magnetisation
- Caused by the nuclei giving up their energy to the surrounding environment or lattice,
- often termed *spin lattice relaxation.*
- The rate of recovery of longitudinal magnetization is an exponential process
- a recovery time constant called T1= time it takes 63% of the longitudinal magnetisation to recover

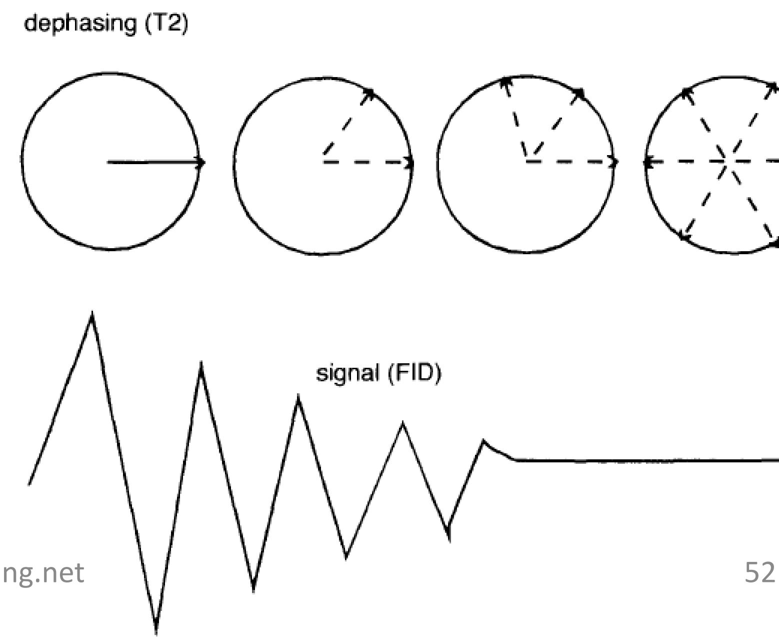


T2 decay

- The decay of transverse magnetisation
- caused by nuclei exchanging energy with neighboring nuclei→ dephasing
- often termed *spin spin relaxation*
- The rate of decay is also an exponential process
- T2 relaxation time of a tissue (time constant of decay): the time it takes 63% of the transverse magnetisation to be lost

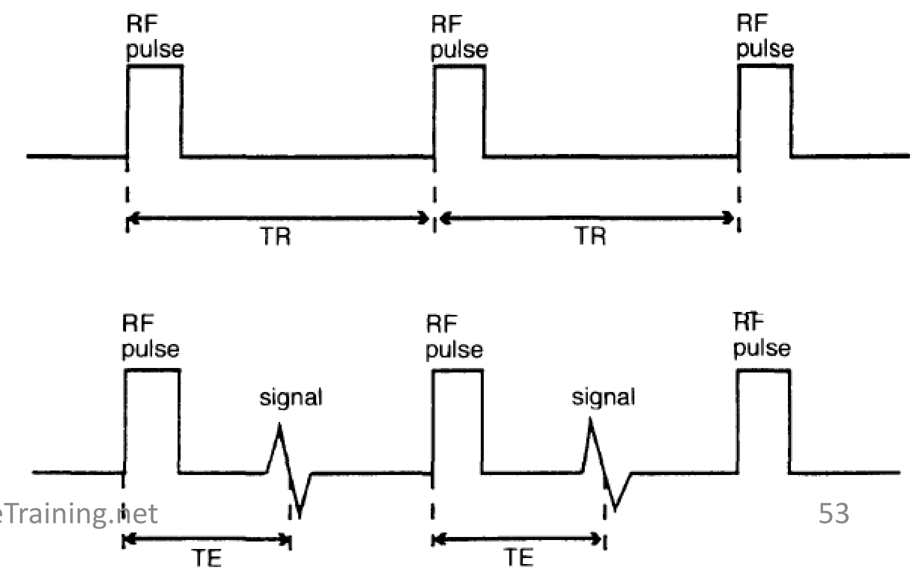
N.B. A signal is only induced in the receiver coil if there is magnetisation in the transverse plane, that is in phase (the signal in the coil here will decays)



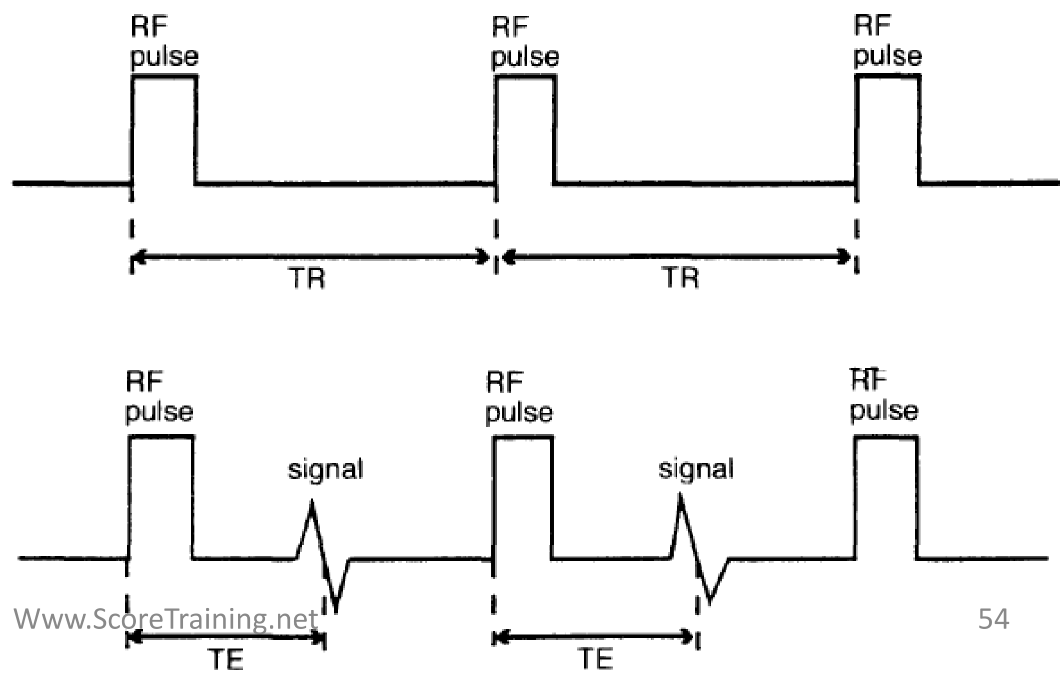


Pulse timing parameters

- A pulse sequence consists of several components:
- **The *repetition time (TR):***
  - *is the time from the application of one RF* pulse to the application of the next RF pulse
  - determines the amount of relaxation that is allowed to occur between the end of one RF pulse and the application of the next.
  - Therefore the TR determines the amount of T1 relaxation that has occurred .



- **The *echo time (TE):***
  - *is the time from the application of the RF pulse to* the peak of the signal induced in the coil
  - The TE determines how much decay of transverse magnetisation is allowed to occur before the signal is read.
  - Therefore, the TE controls the amount of T2 relaxation that has occurred.

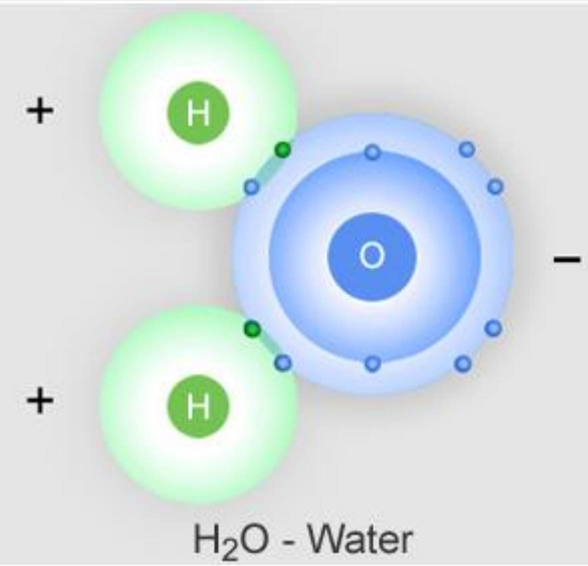


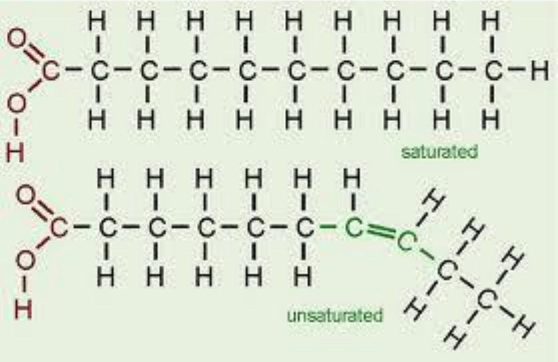
Fat and water hydrogen MRI characteristics

Fat and water differences

1- Larmor frequency

- **Water**
  - is hydrogen linked to oxygen which tends to steal the electrons away from around the hydrogen nucleus. → H is more available to the effects of the main magnetic field.





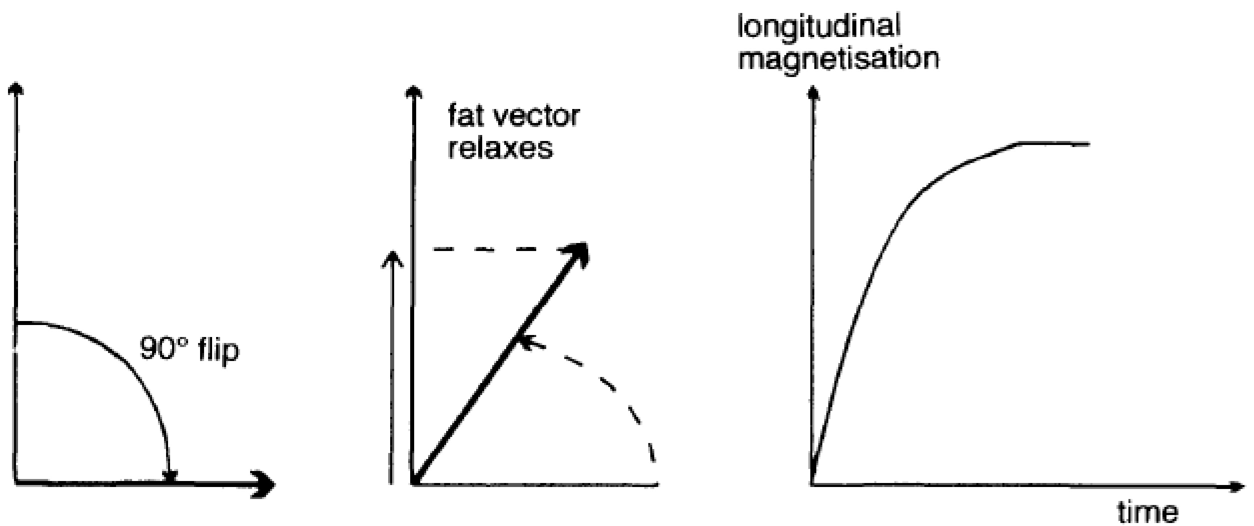
- **Fat**
  - is hydrogen linked to carbon ,which does not take the electrons from around the hydrogen nucleus → electron cloud protect the nucleus from the effects of the main field.
- **Results**:
  - Larmor frequency of hydrogen in water is higher than hydrogen in fat.

2- T1 recovery

– Fat

- Slow molecular tumbling in fat → T1 recovery process is relatively rapid. →magnetic moments of fat nuclei regain their longitudinal magnetisation quickly.
- Result:

T1 time of fat is short (100-150 ms)

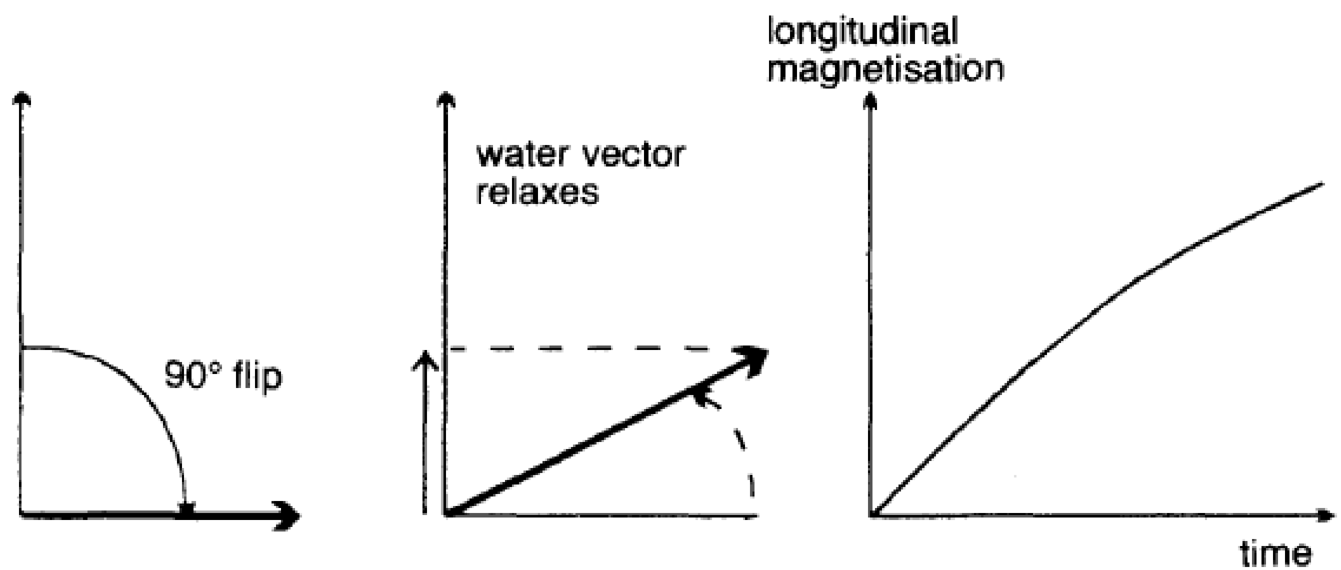


fat regains longitudinal magnetisation rapidly

– *Water*

- In water, molecular mobility is high resulting in less efficient T1 recovery→ The magnetic moments take longer to relax and regain their longitudinal magnetisation.
- Result:

T1 time of water is long (1500-2000 ms)



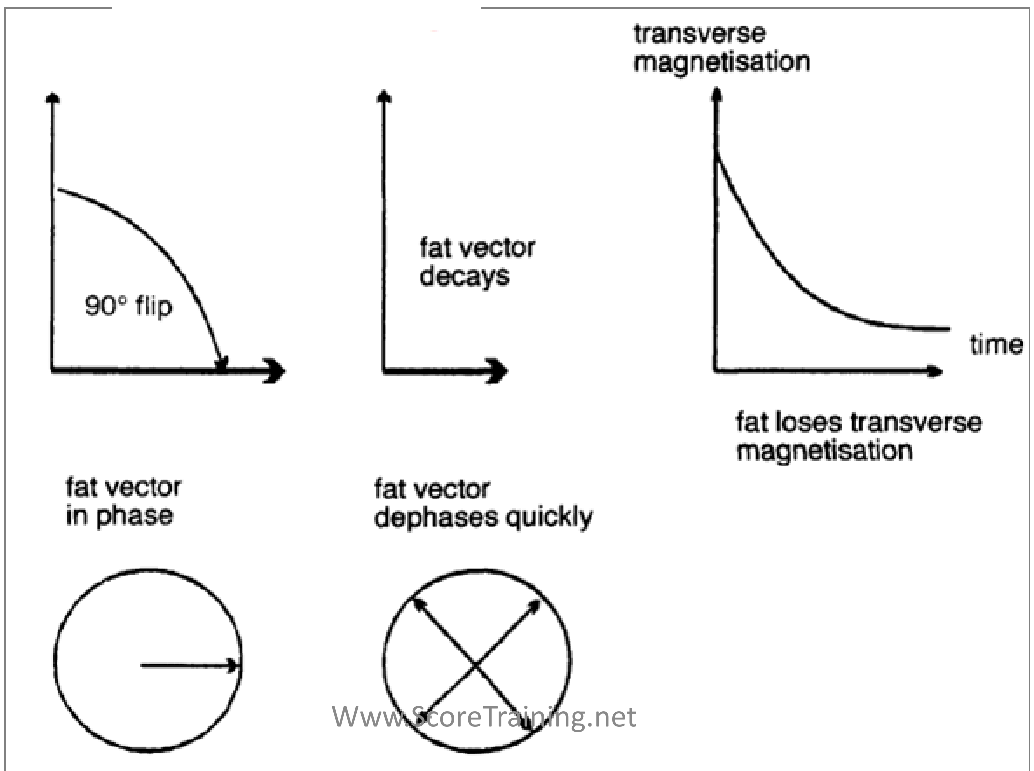
water regains longitudinal magnetisation slowly

<u>3- T2 *decay*</u>

– *Fat*

- energy exchange between nuclei is more efficient in the hydrogen in fat → more rapid dephasing
- Result:

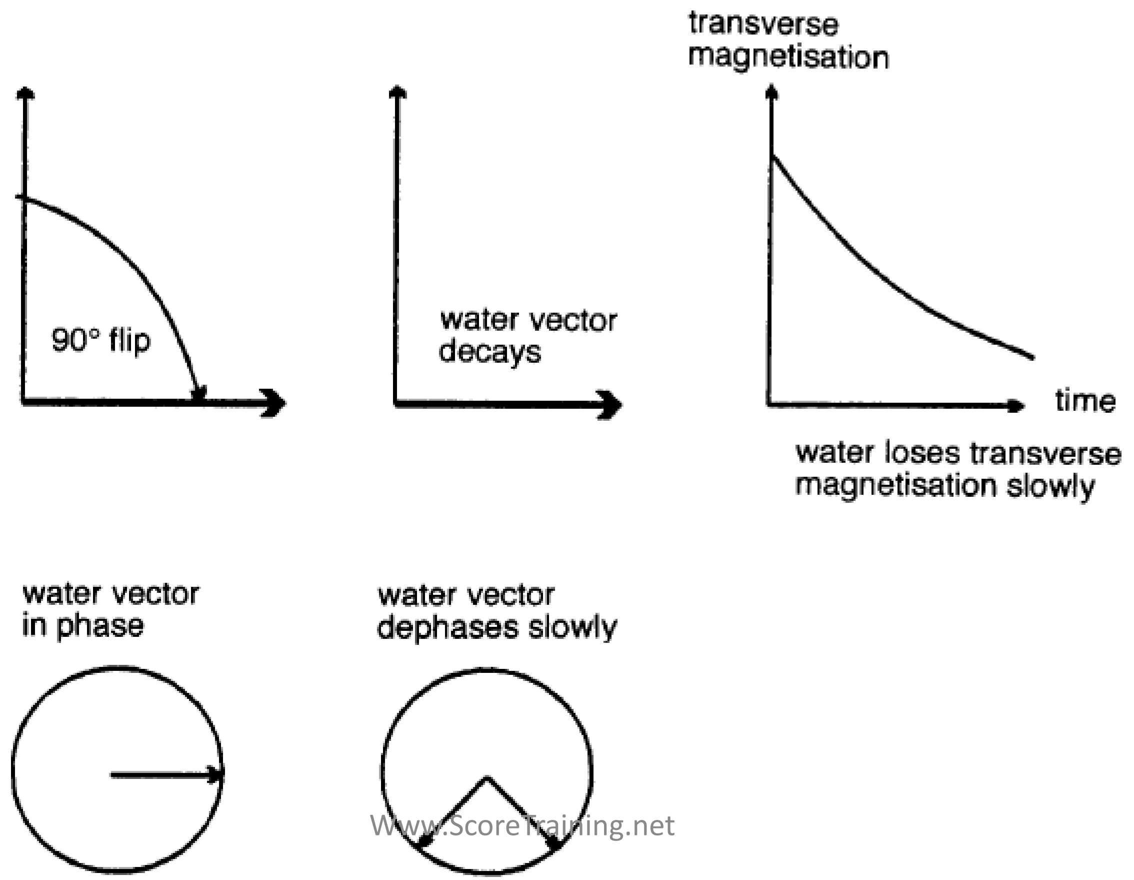
T2 time of fat is short (80ms.)



– ***water***

- energy exchange in water is less efficient than in fat→ slow dephasing
- Result:

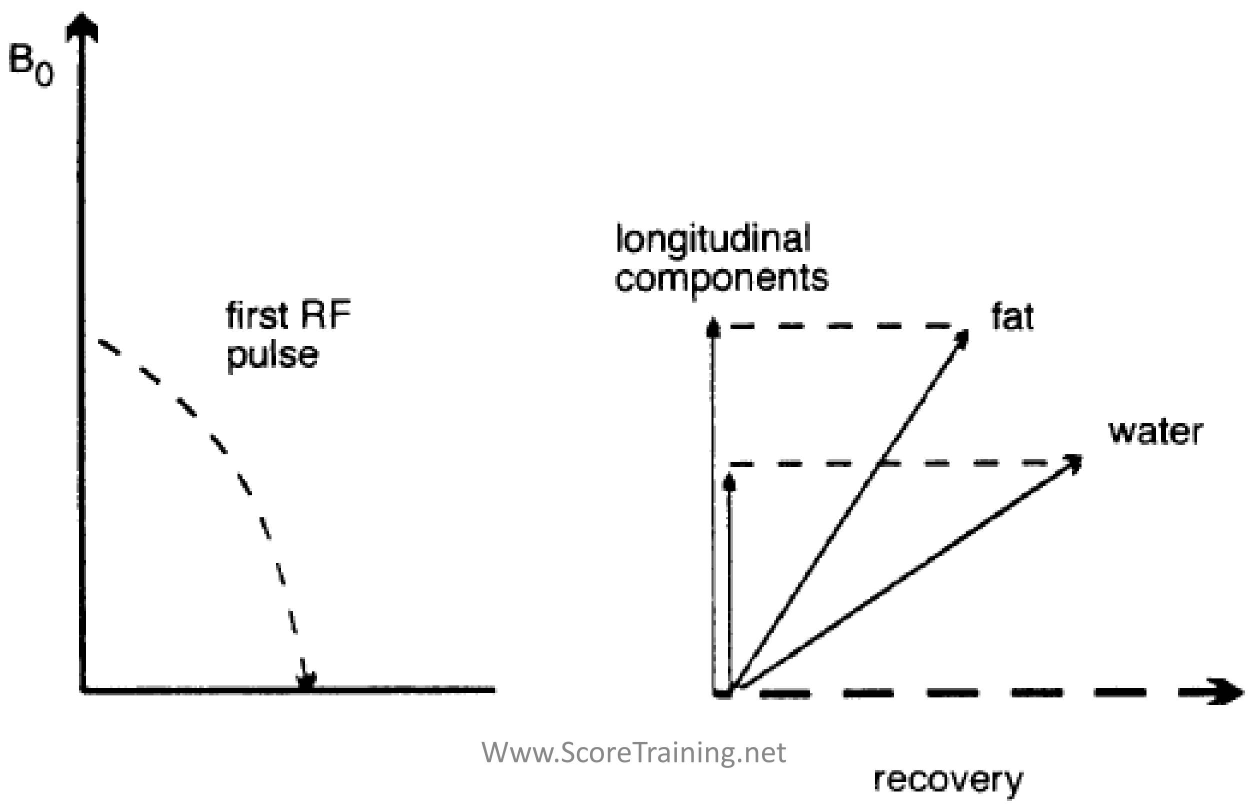
T2 time of water is long (200 ms)



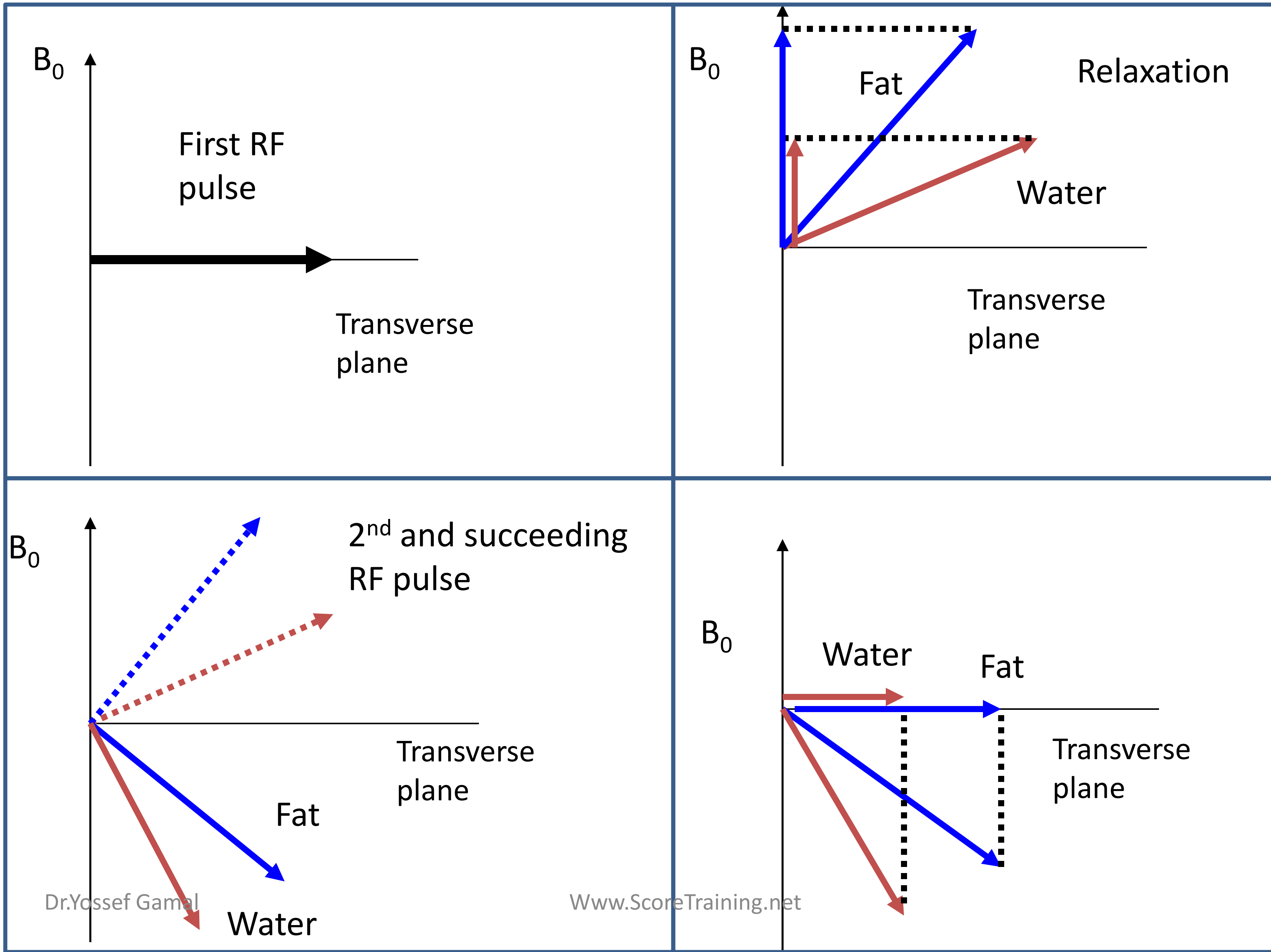
Causes of contrast between tissues in MRI images

T1 contrast

- T1 time of fat is shorter than water→ the fat vector realigns with Bo faster than water → The longitudinal component of magnetization of fat is larger than water at a given time.

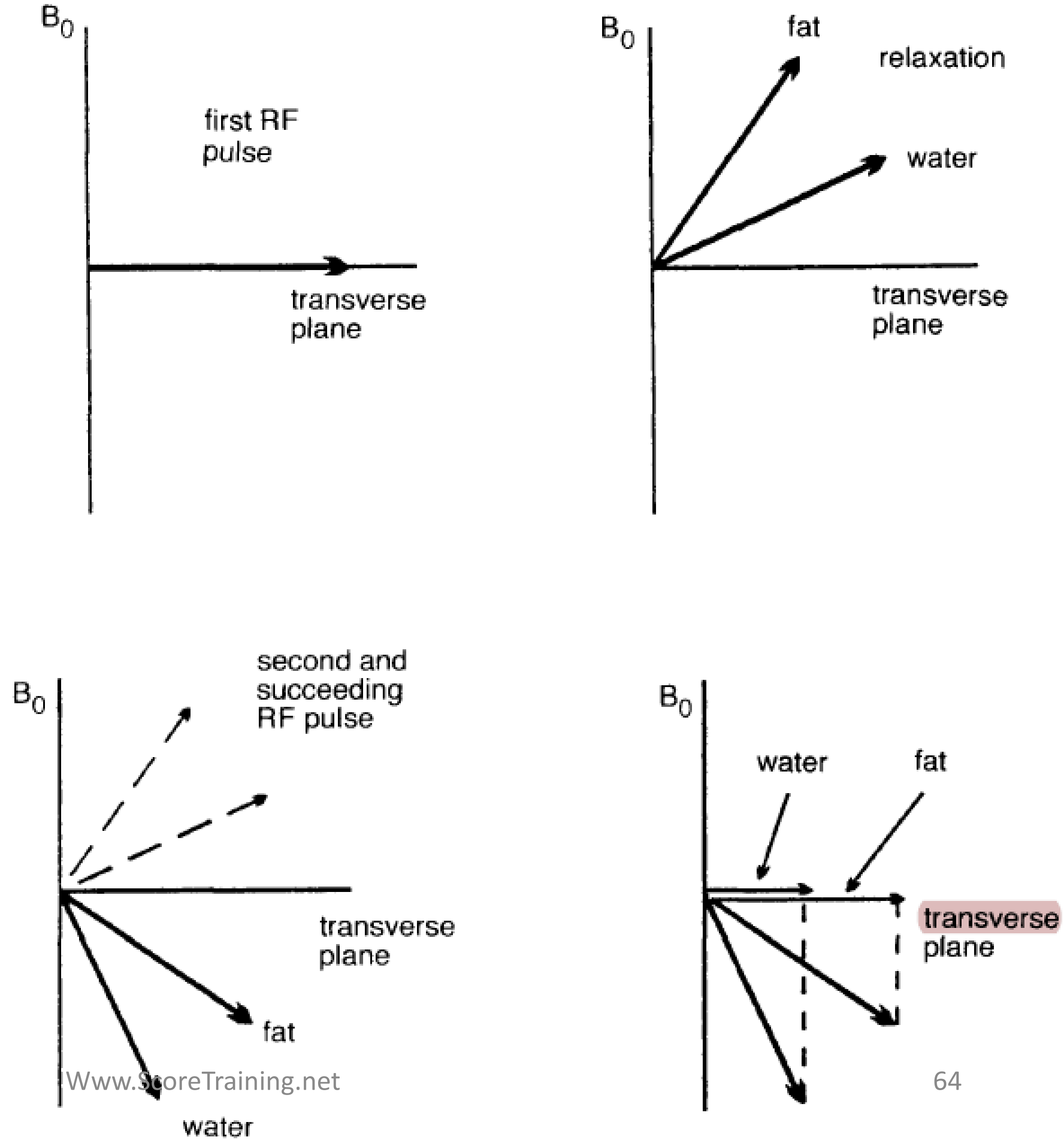


- After a certain TR, the next 90° RF excitation pulse is applied →the longitudinal components of magnetisation of both fat and water into the transverse plane are flipped



= saturation:

- i.e. After the 2nd 90° RF NMV is pushed beyond 90° (short TR)
- it is said to be ***partially saturated → T1 wt image***



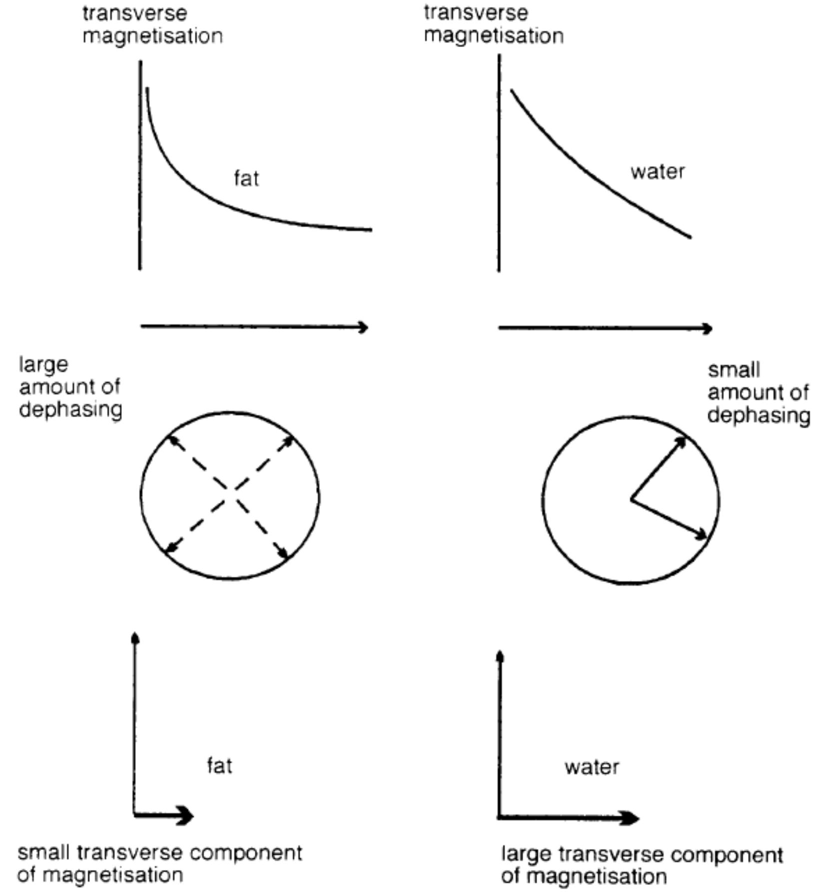
- As there is more longitudinal magnetisation in fat before the RF pulse → there is more transverse magnetisation in fat after the RF pulse → Fat therefore has a high signal and water has a low signal on a T1 contrast image.
- Such images are called: **T1 weighted images.**

T2 contrast

- T2 time of fat is shorter than that of water→ transverse component of magnetisation of fat decays faster

i.e. at a given time magnitude of transverse magnetisation in water is larger

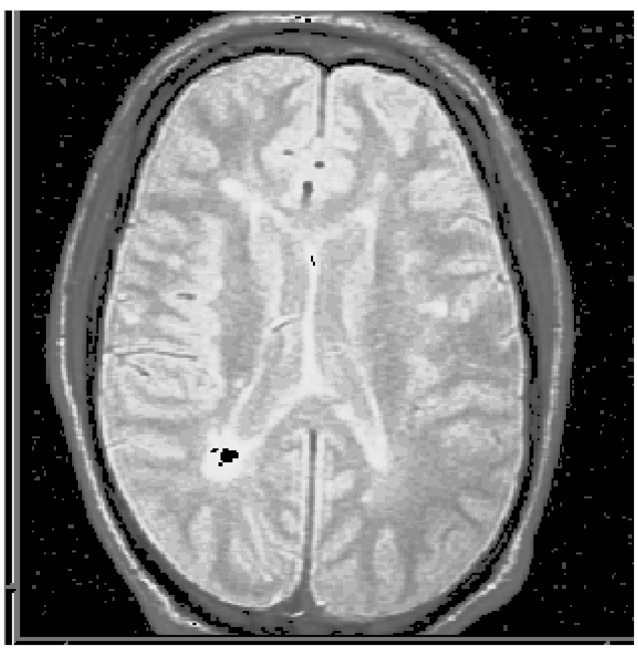
- Result:
  – After TE when signal is received: Water has a high signal and Fat therefore has a low signal
- Such images are called: **T2 weighted images.**

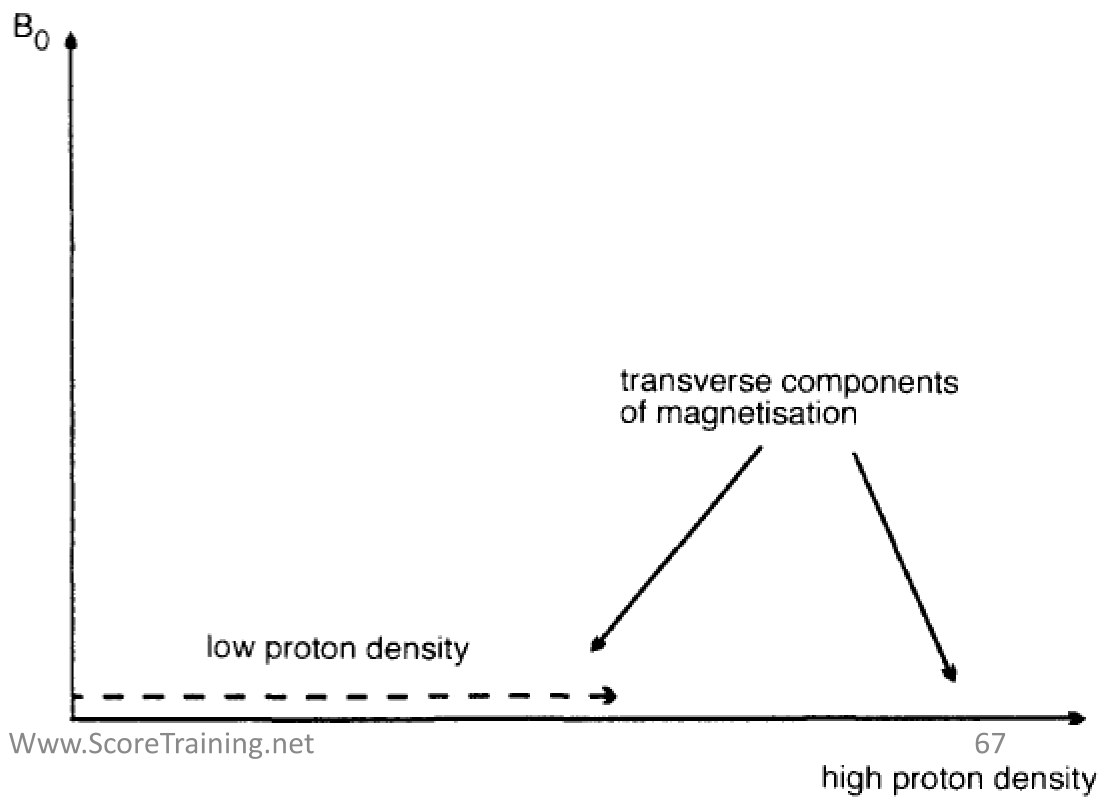


Proton density contrast

definition: differences in signal intensity between tissues due to their relative number of protons per unit volume.

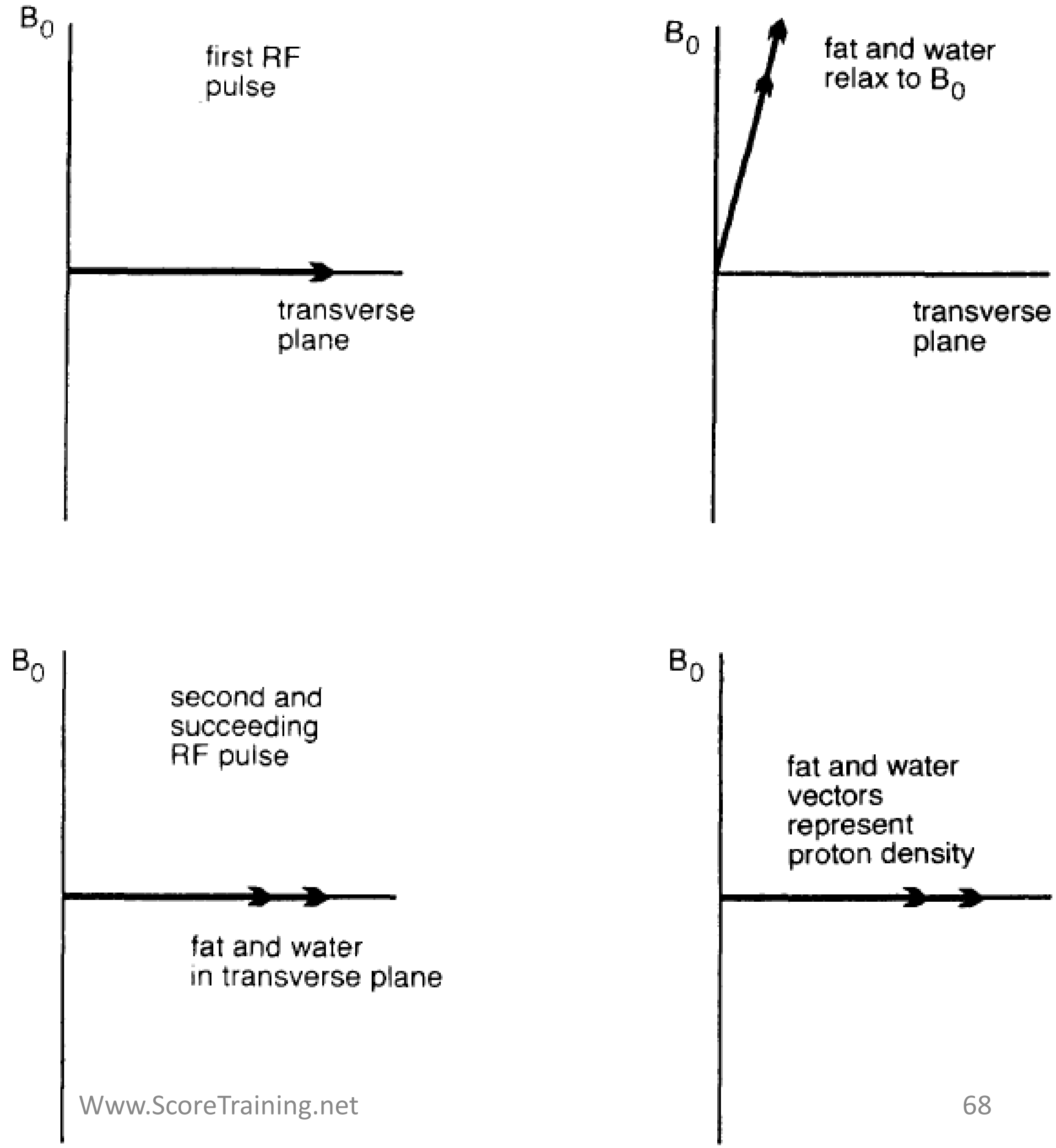
- Tissues with high proton density (e.g. brain) have a large transverse component of magnetisation (high signal)
- Tissues with low proton density (e.g. cortical bone) have a small transverse component of magnetisation (low signal)
- Proton density contrast is always present





= non saturation:

- after long TR, second RF will push fat and water NMV to 90° it is said to be ***not saturated → PD wt image*** (The magnitude of the transverse component depends only on their individual proton densities)



MRI image weighting

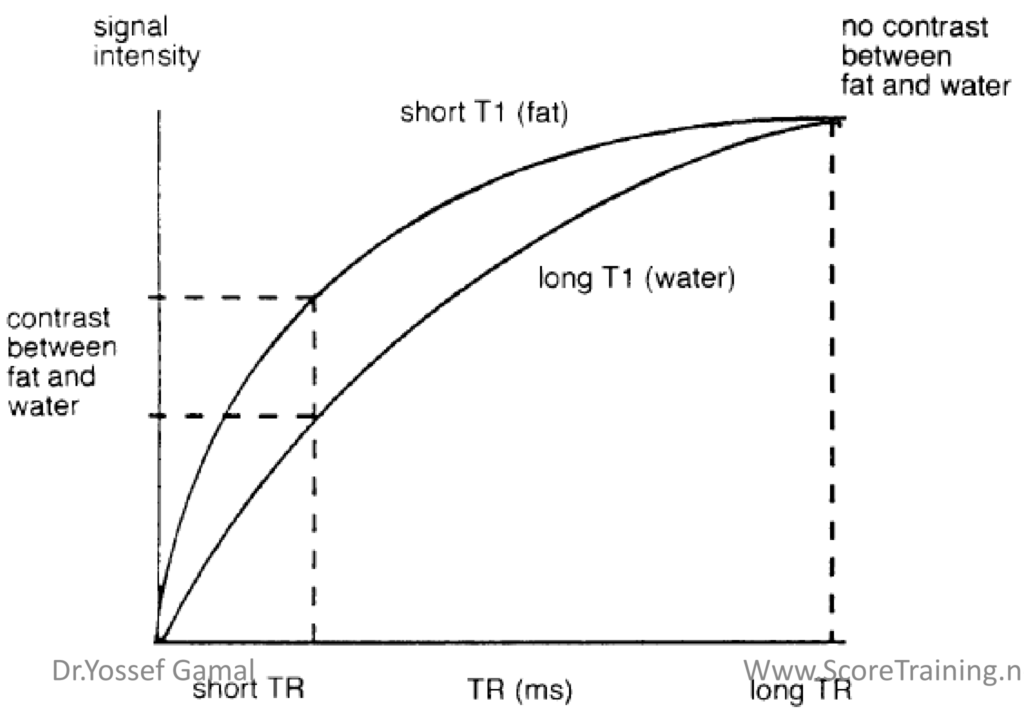
Weighting:

The selection of appropriate TR and TE so that one contrast mechanism predominates over the other two.

T1 *weighting*

T1 weighted image: image where the contrast depends on the differences in the T1 times between fat and water

- To achieve T1 weighting: the TR must be short enough so that neither fat nor water has sufficient time to fully return to Bo.
- If the TR is too long, both fat and water return to Bo fully → differences in their T1 times are not demonstrated.



Important facts:

- TR controls the amount of T1 weighting.
- For T1 weighting the TR must be short.

T2 weighting

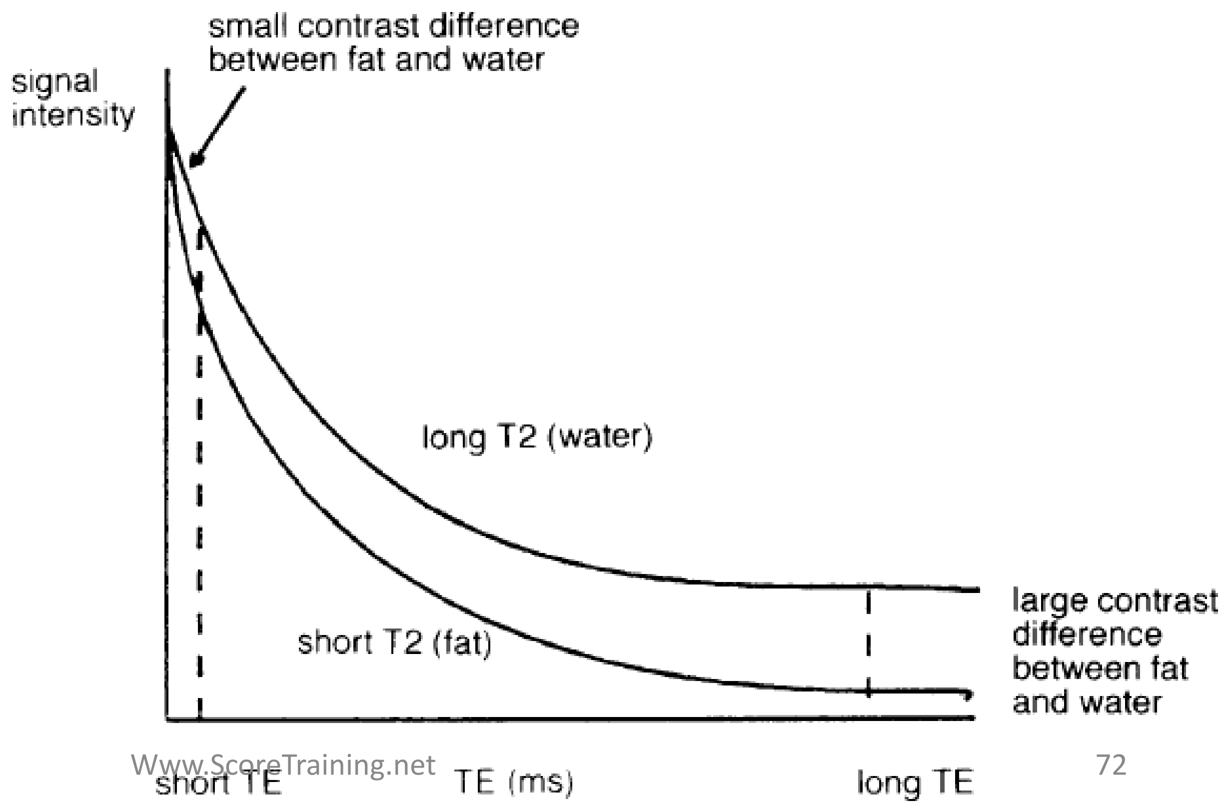
T2 weighted image: image where the contrast predominantly depends on the differences in the T2 times between fat and water

- To achieve T2 weighting: the TE must be long enough to give both fat and water time to decay.
- If the TE is too short, neither fat nor water has had time to decay, and therefore the differences in their T2 times are not demonstrated in the image

Important facts

• TE controls the amount of T2 weighting.

• For T2 weighting the TE must be long.



Proton *density weighting*

proton density image: image where the difference in the numbers of protons per unit volume is the main determining factor in forming image contrast.

- In order to achieve proton density weighting, the effects of T1 and T2 contrast must be diminished, so that proton density weighting can dominate.
- i.e. :
  - long TR (allows both fat and water to fully recover their longitudinal magnetisation→ diminishes T1 weighting)
  - short TE (does not give fat or water time to decay →diminishes T2 weighting)

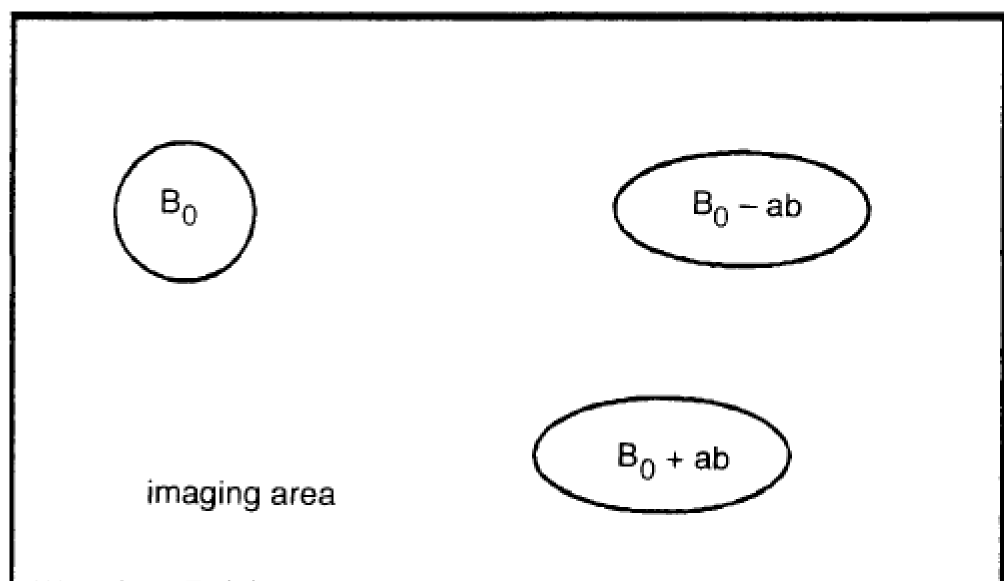
Typical values of TR and TE

Long TR	**2000 ms+**
Short TR	**250-700 ms**
Long TE	**60 ms+**
Short TE	**10-25 ms**

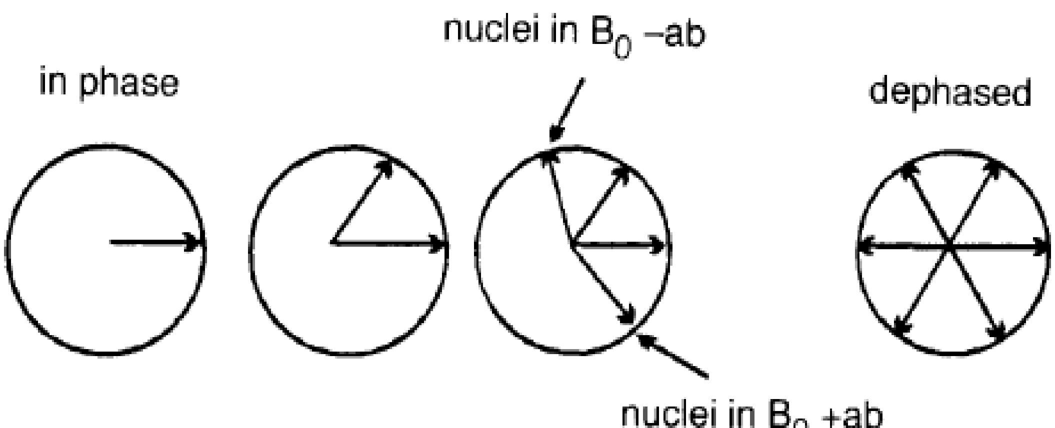
T2* decay

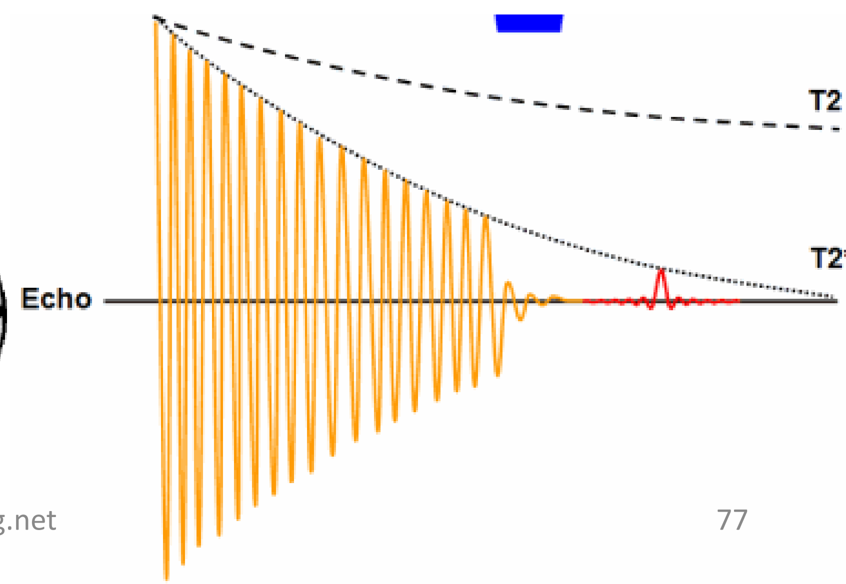
T2* decay

- T2* decay is the decay following removal RF excitation pulse (FID).
- This decay is **faster** than T2 decay since it is a combination of two effects:
  - – (1) T2 decay itself
  - – (2) dephasing due to magnetic field inhomogeneities.



- **Effect of magnetic field inhomogeneities:**
  - nucleus in an area with a higher field strength→ precessional frequency increases (larmor equation)
  - nucleus in an area of inhomogeneity with a lower field strength →precessional frequency decreases
  - This relative acceleration and deceleration → immediate dephasing of the NMV

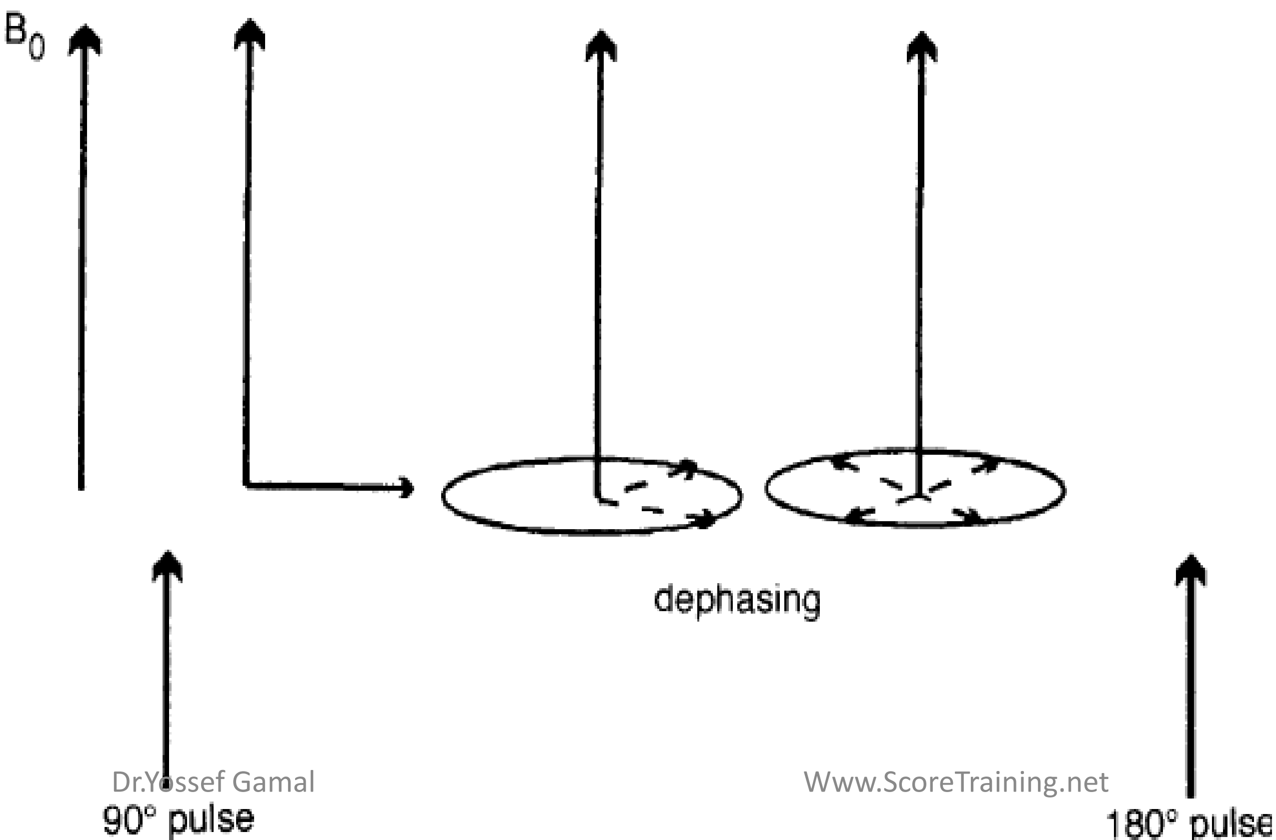


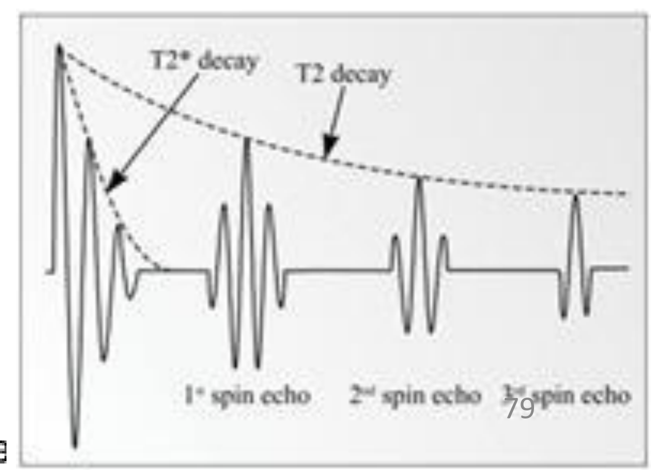


The conventional spin echo pulse sequence

The spin echo pulse sequence

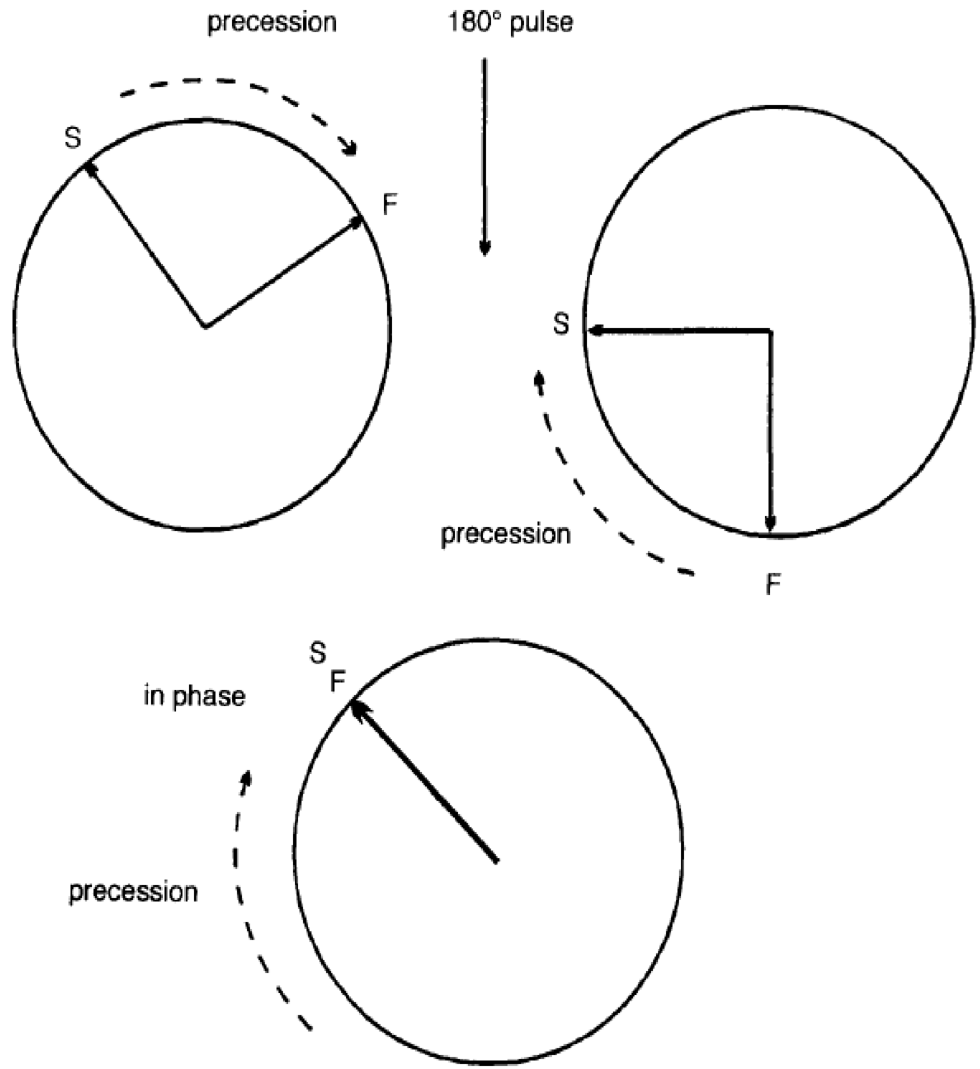
- The spin echo pulse sequence utilizes a 90^o excitation pulse to flip the NMV into the transverse plane.
- When the 90° RF pulse is removed T2* dephasing occurs immediately, and the signal decays.
- **<u>A 180° RF pulse is then used to compensate for the dephasing caused by the external magnetic field inhomogeneities</u>**





Mechanism of action of 180° RF pulse:

- The T2* dephasing causes the magnetic moments to dephase or 'fan out' in the transverse plane.
- The magnetic moments that slow down, form the trailing edge of the fan ,and the magnetic moments that speed up, form the leading edge of the fan



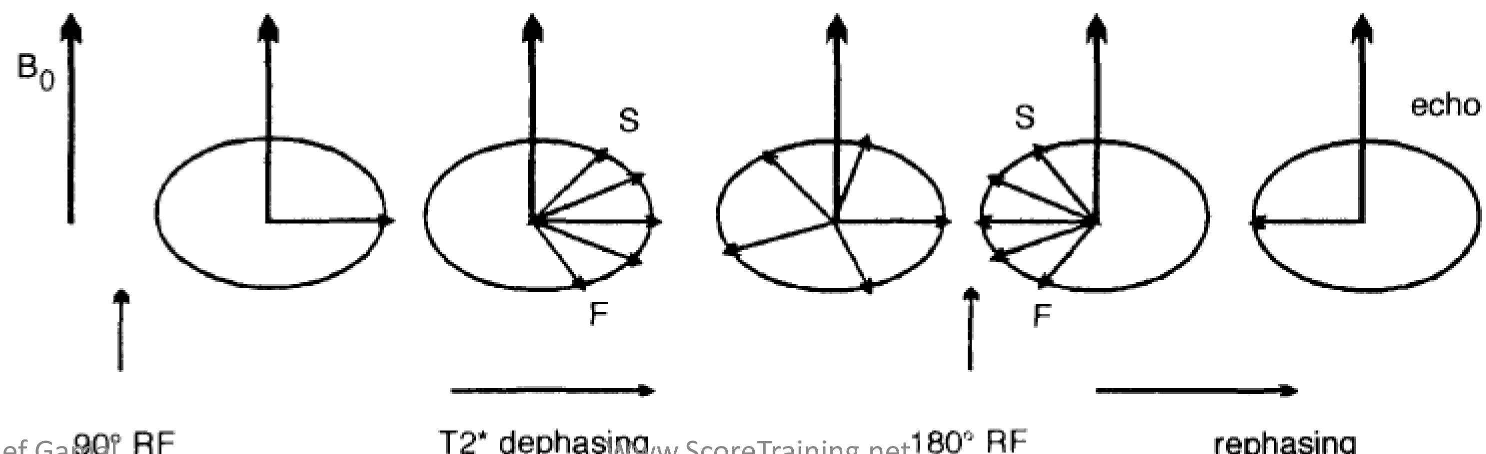
- **The 180° pulse move the NMV through 180°.**

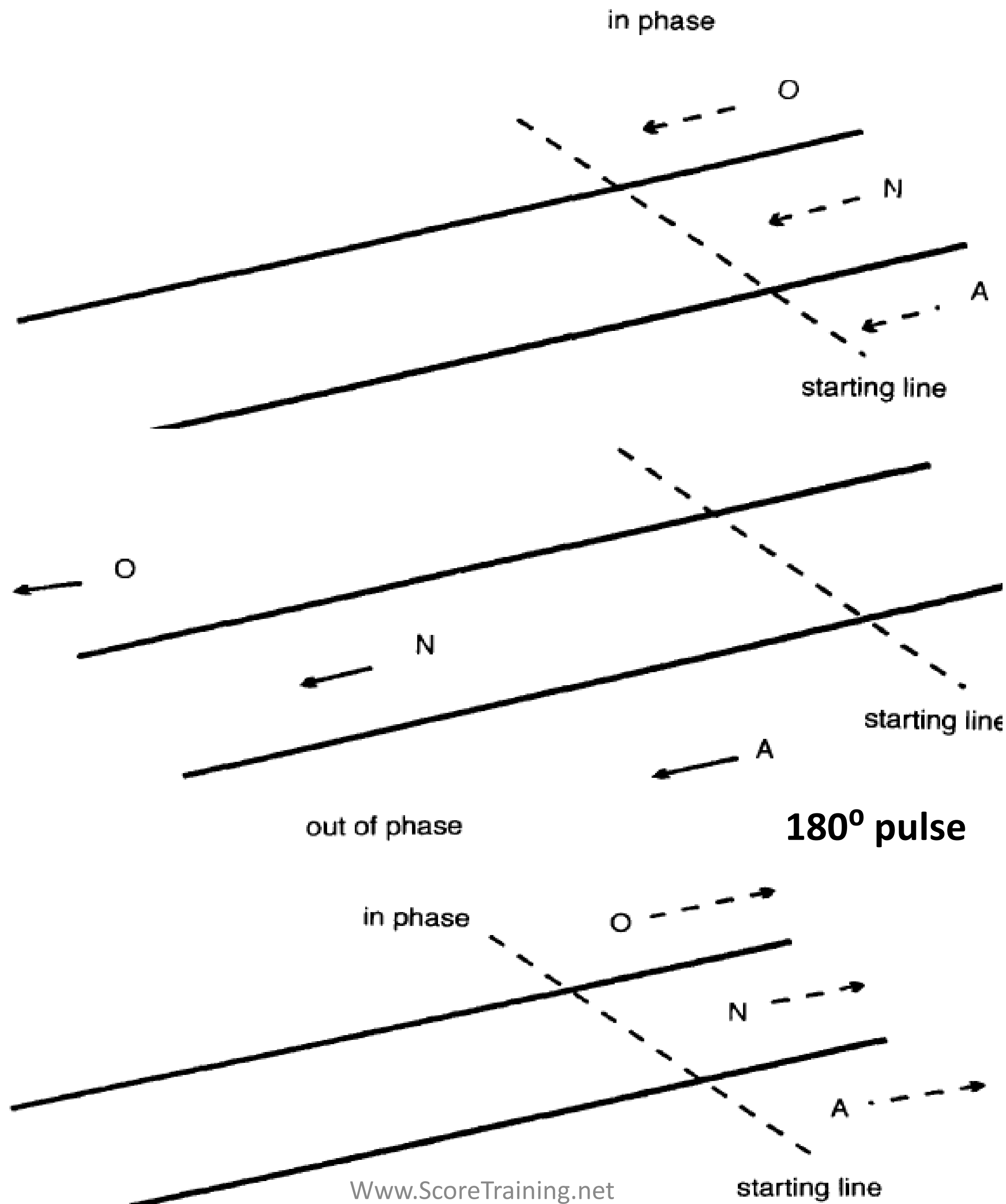
→ the magnetic moments that formed the trailing edge before the 180° pulse, form the leading edge. & the magnetic moments that formed the leading edge prior to the 180° pulse, now form the trailing edge.

- **After removal of 180° pulse:**

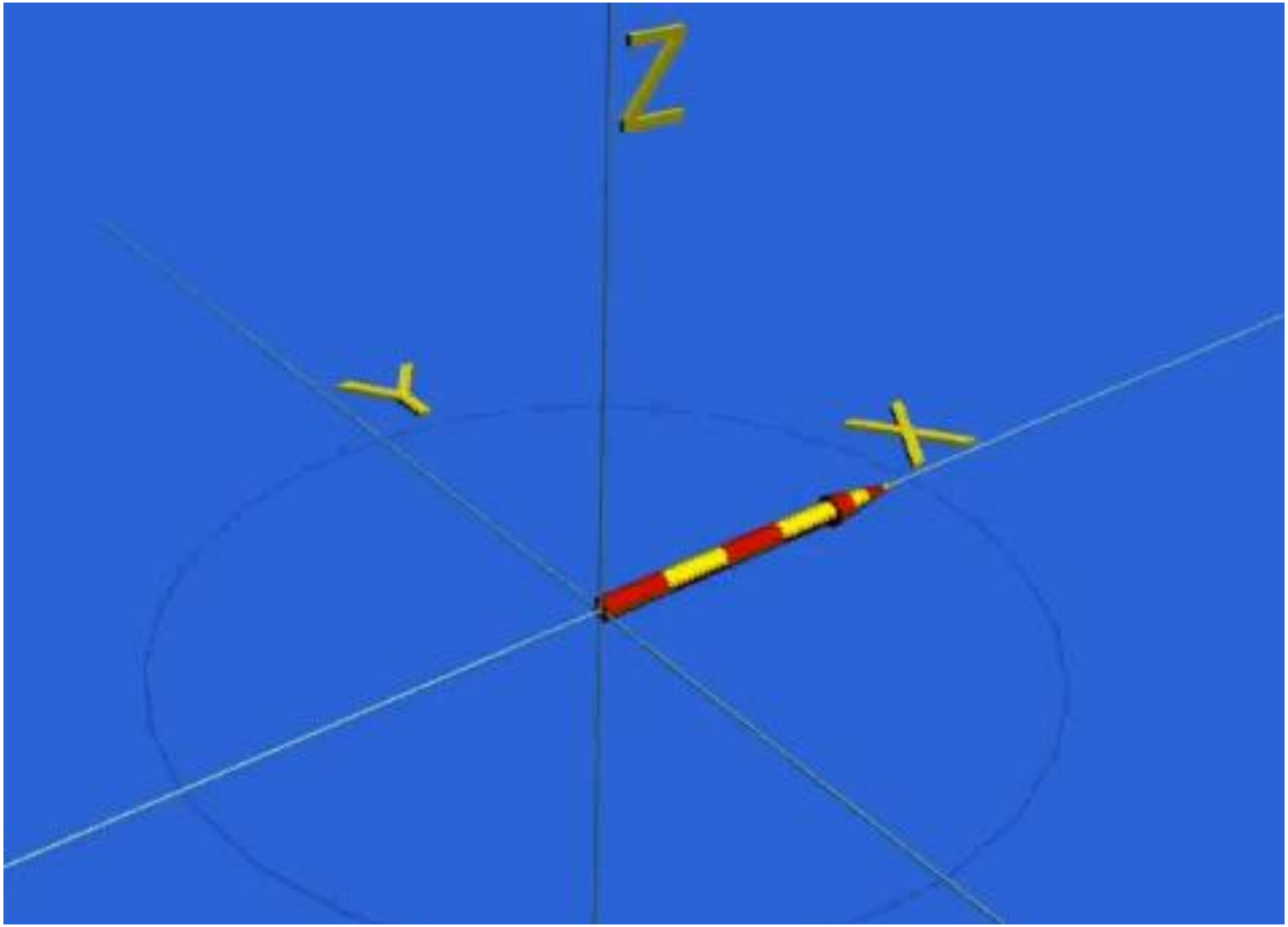
→ the trailing edge begins to catch up with the leading edge. & At a specific time later, the two edges are superimposed. The magnetic moments are now in phase → maximum signal is induced in the coil.

- ***This signal is called a*** ***spin echo*** (contains T1 and T2 information , T2* dephasing has been reduced)



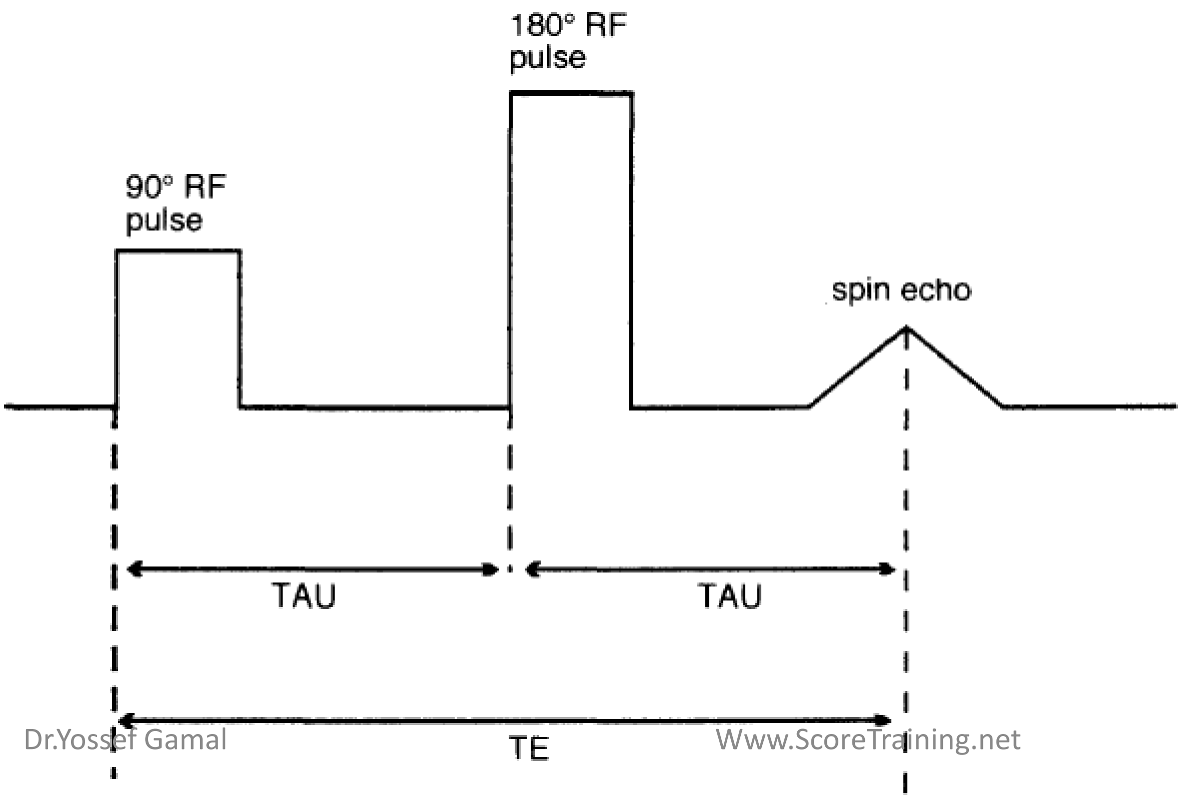

in phase
O
N
A
starting line
O
N
starting line
A
out of phase
in phase
O
N
A
starting line

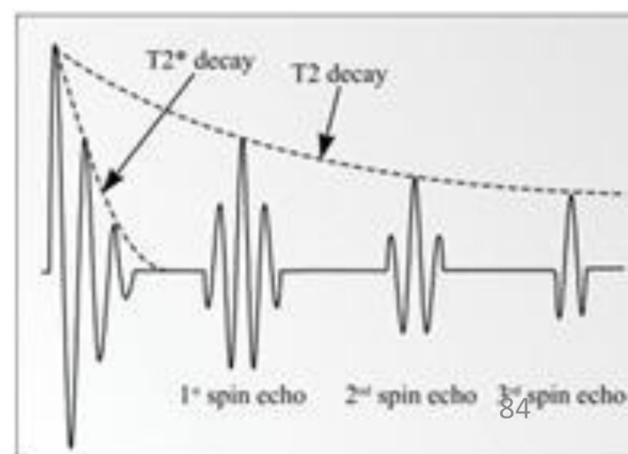
180^o pulse


Z
Y
X

Timing parameters in spin echo

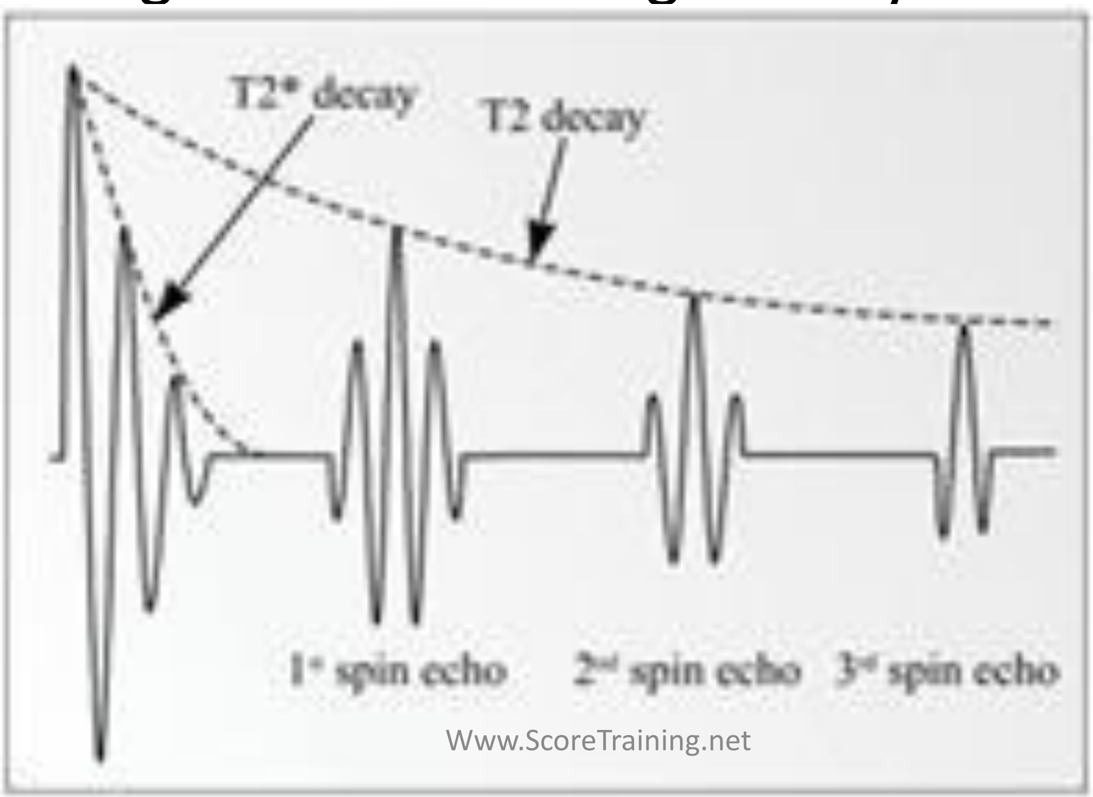
- **TR:**
  - the time between each 90^o excitation pulse.
- **TE:**
  - the time between the 90^o excitation pulse and the peak of the spin echo.
  - = the time the NMV took to dephase when the 90^o RF pulse was withdrawn **+** The time taken to rephase after the application of the 180^o RF pulse (both are equal = *TAU time)*
  - *TE is therefore twice the TAU*





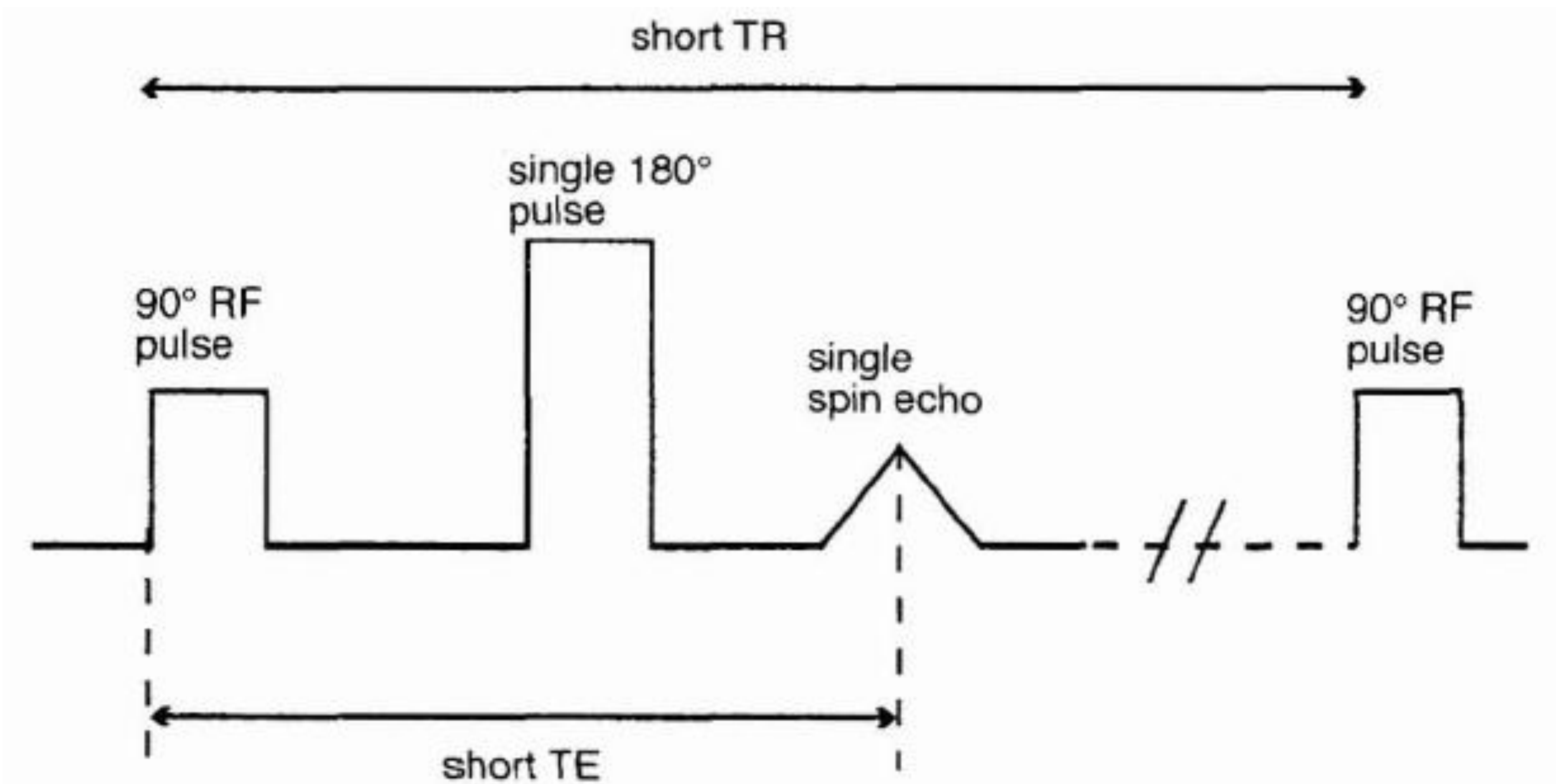
Important note

- In most spin echo pulse sequences, more than one 180^{o} RF pulse can be applied after the 90^{o} excitation pulse.
- Each 180^{o} pulse generates a separate spin echo that can be received by the coil and used to create an image.
- each time the echo is less due increasing TE so that the transverse magnitisation is lost gradually



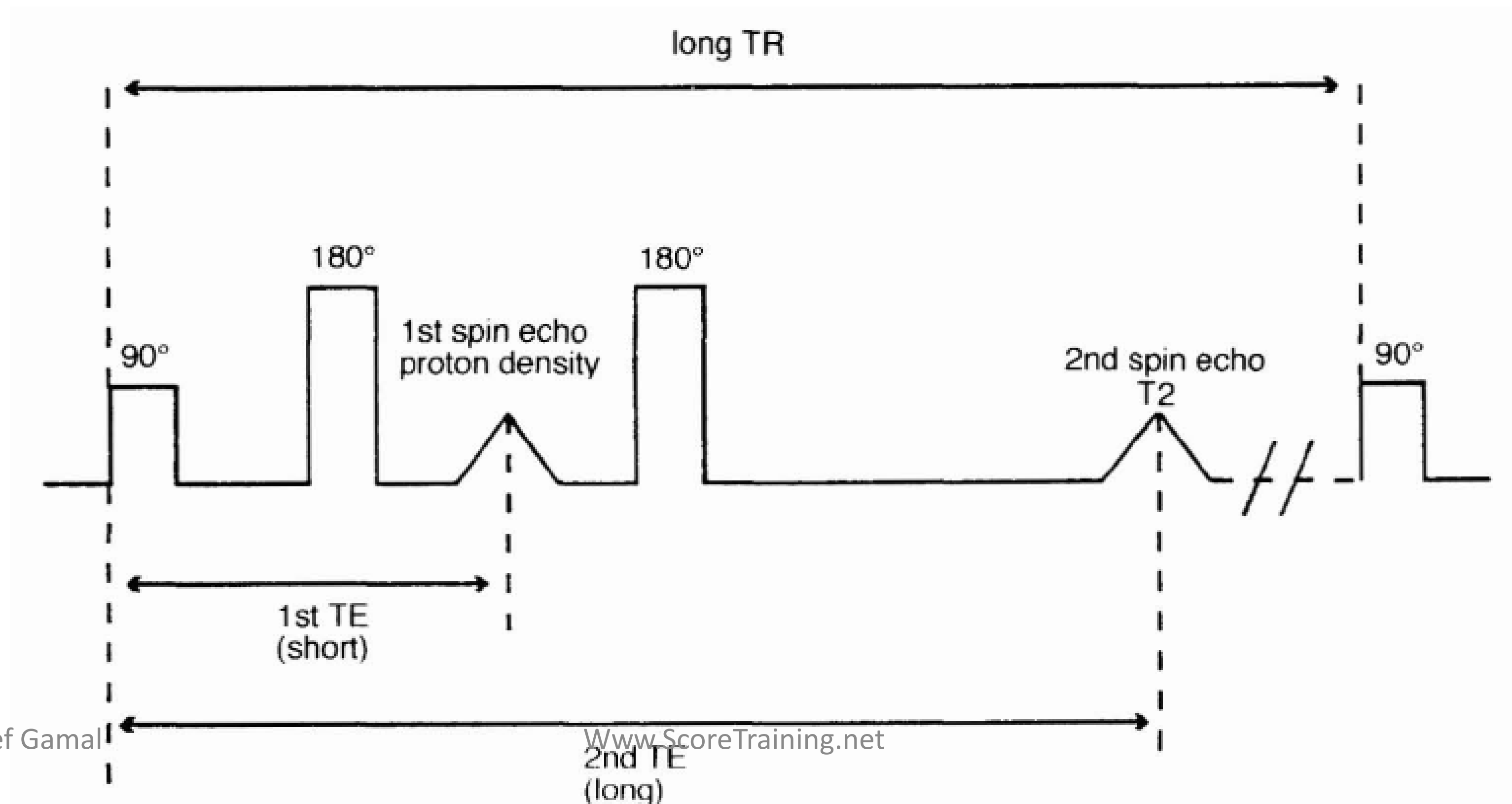
Spin echo sequence using one echo

- This pulse sequence can be used to produce T1 weighted images if a short TR and TE are used



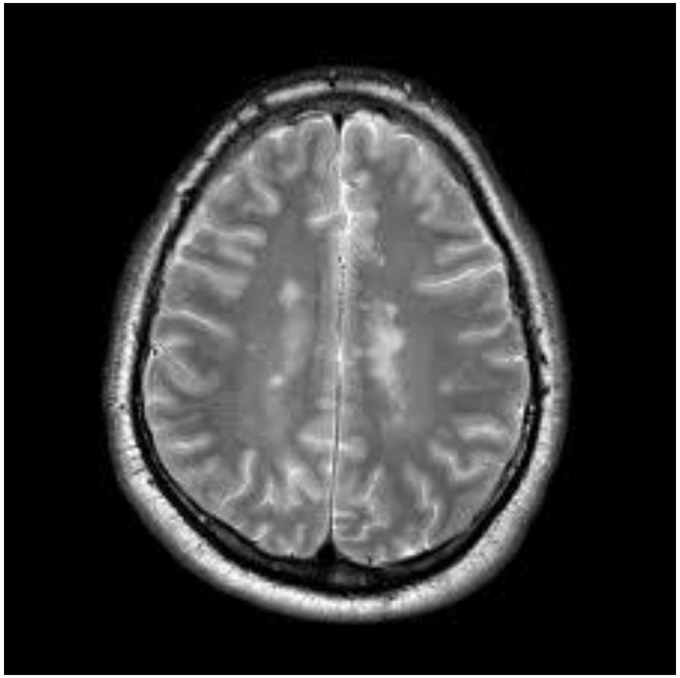
Spin *echo using two echoes*

- This can be used to produce both a proton density and a T2 weighted image
- Long TR is selected
- The first spin echo is generated early by selecting a short TE (PD)
- The second spin echo is generated much later by selecting a long TE (T2WI).



Uses of spin echo pulse

- the gold standard for most imaging.
- T1 weighted images are useful for demonstrating anatomy
- T2 weighted images also demonstrate pathology. ***Parameters***



Advantages

- Good image quality

Disadvantages

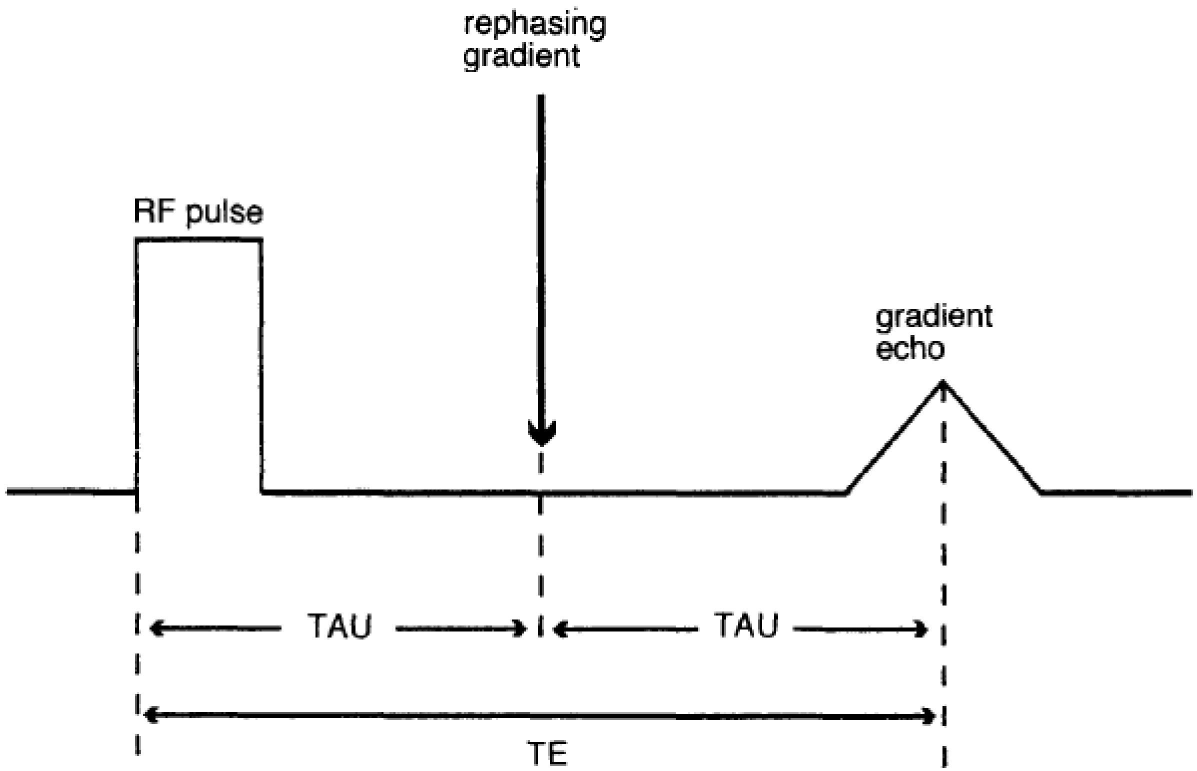
- Scan times relatively long

summary

- A spin echo pulse sequence uses a 90° excitation pulse followed by one or more 180° rephasing pulses to generate one or more spin echoes.
- Spin echo pulse sequences produce either Tl, T2 or proton density
- TR controls the T1 weighting.
- TE controls the T2 weighting.

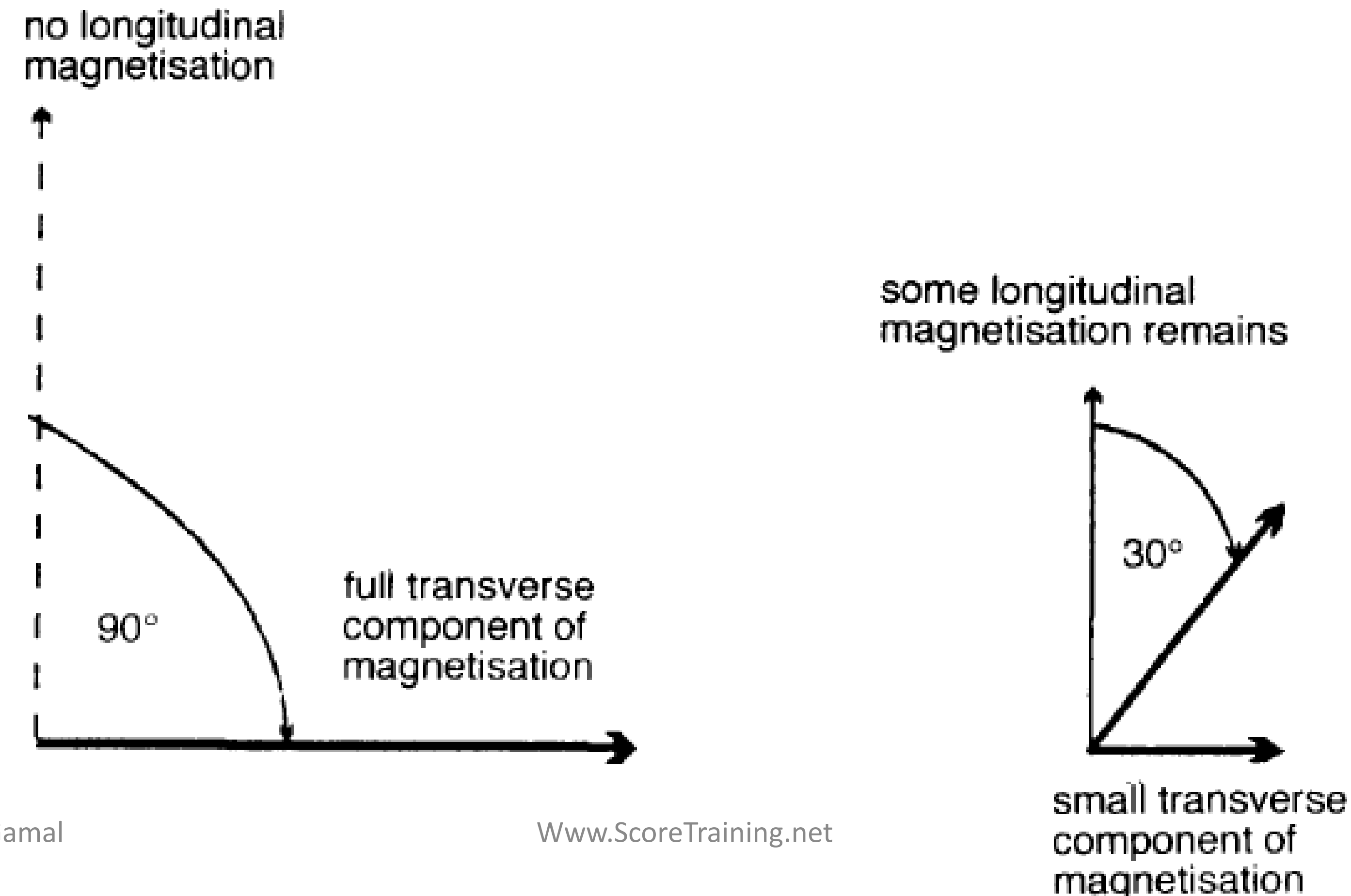
The gradient echo pulse sequence

gradient echo pulse sequence



gradient echo pulse sequence

- Differences from spin echo sequence:
  - Utilizes RF excitation pulse other than 90°
  - the magnitude of the transverse component created is less than in spin echo (where all the longitudinal magnetisation is converted to transverse plane)
  - The magnetic moments are then rephased by a *gradient (not 180°pulse).*
  - signal received by the coil is called a *gradient echo.*



Magnetic field Gradients

Definition:

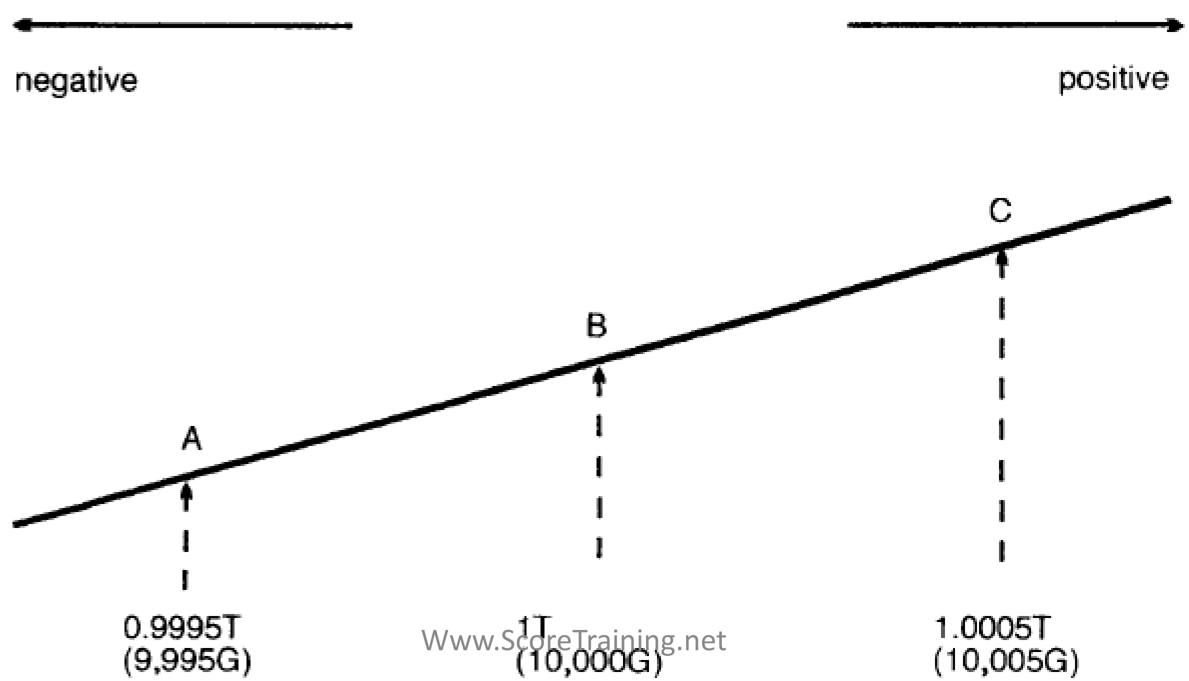
- alterations to the main magnetic field

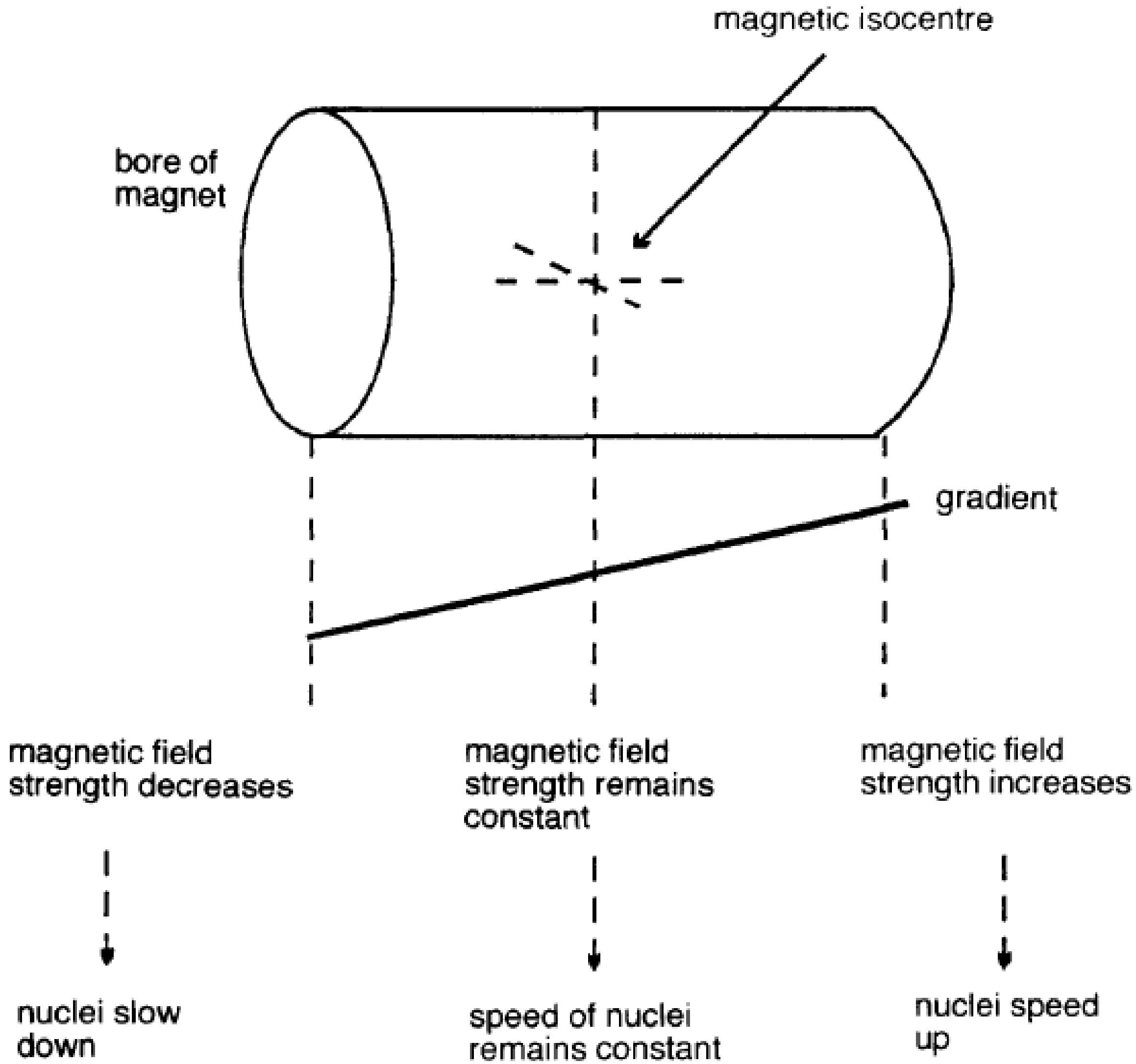
Source:

- generated by coils of wire located within the bore of the magnet through which current is passed → induces a magnetic field around it, that either subtracts from, or adds to Bo.
- The magnitude of Bo is altered in a linear fashion → magnetic field strength and precessional frequency experienced by nuclei situated along the axis of the gradient is changed . This is called spatial encoding .

Function:

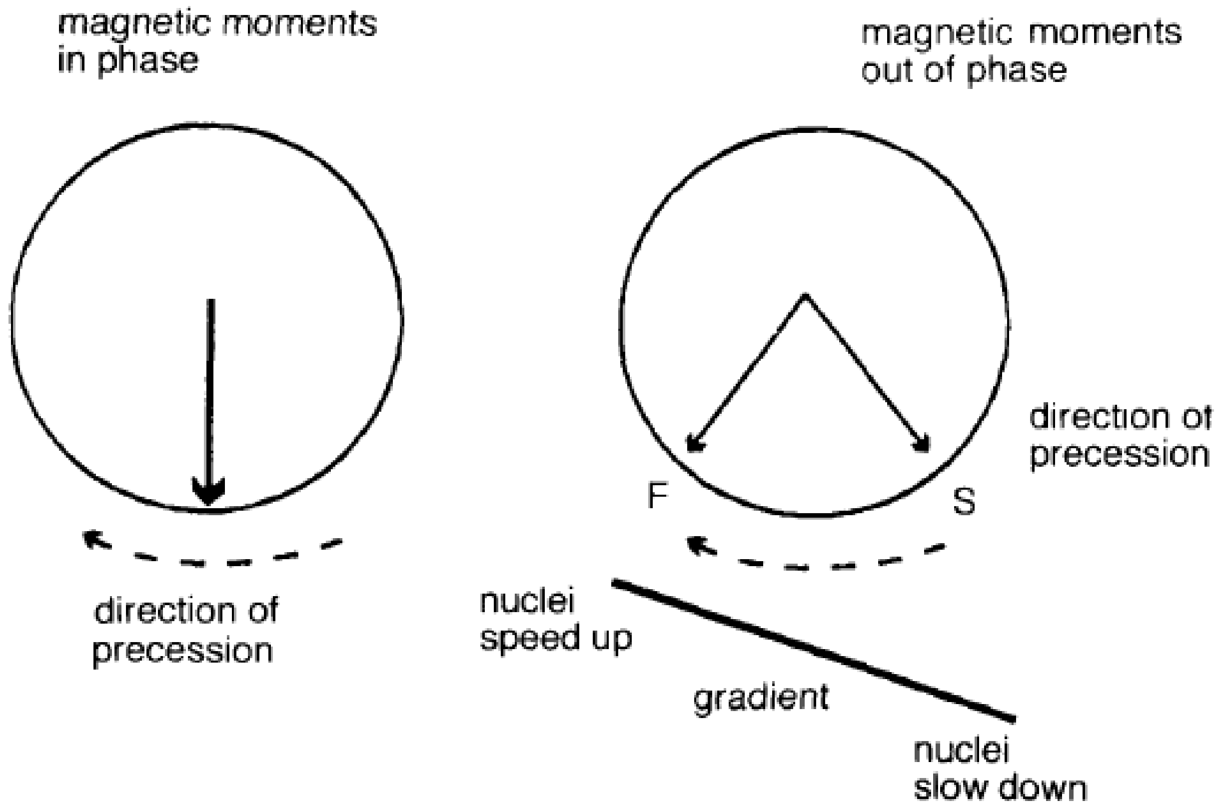
- Used to rephase the nuclei (after gradient dephasing)
- Other uses discussed later



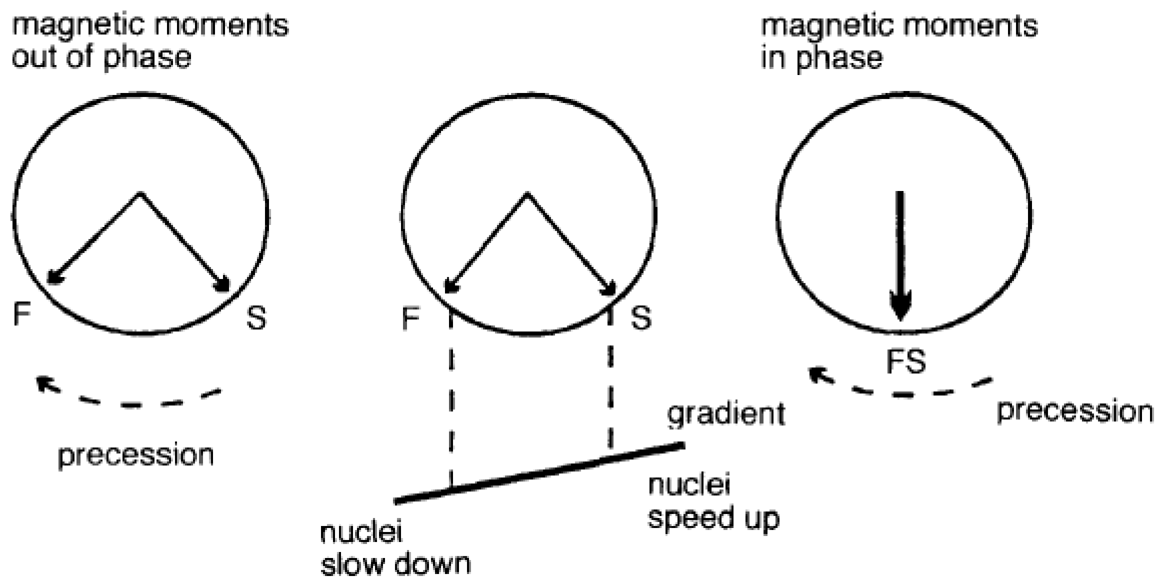


– As gradients can cause nuclei to speed up or slow down, they are used to *dephase and rephase their magnetic moments.*

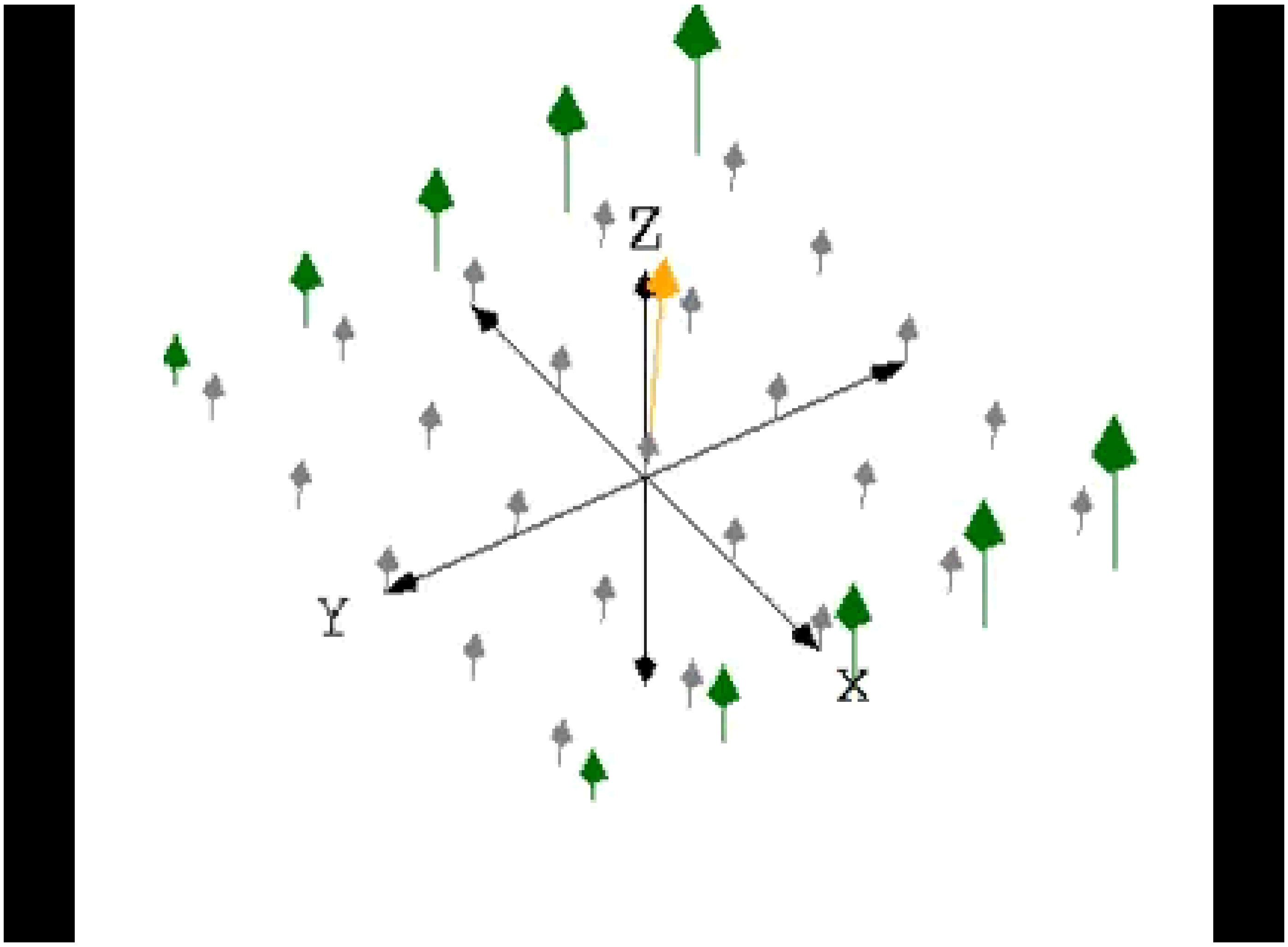
How gradients dephase



How gradients rephase



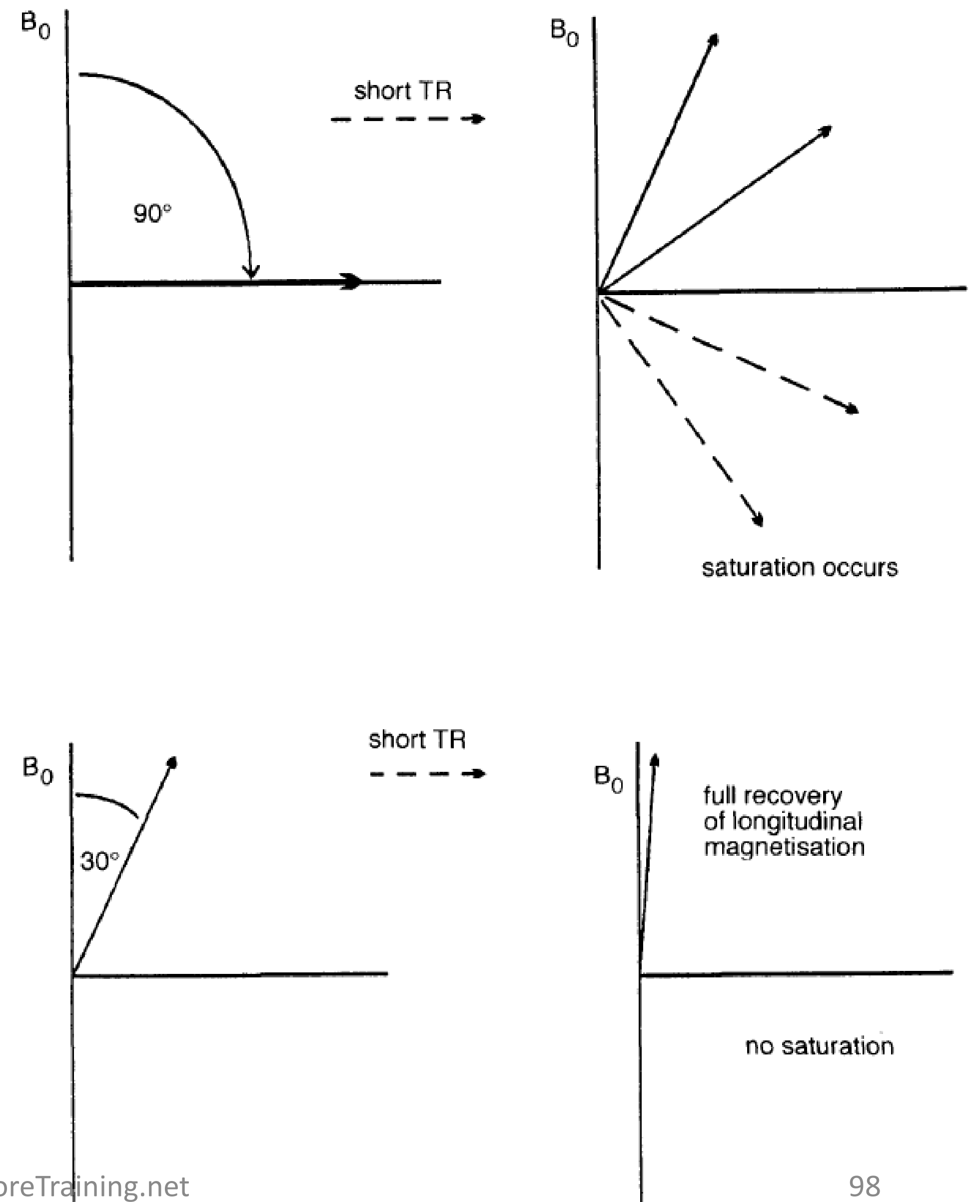
- As the nuclei are rephased by the gradient. A maximum signal is induced in the receiver coil


Z
Y
X

advantages of gradient echo sequences

1) gradients can rephase faster than 180º RF pulses → minimum TE is shorter than in spin echo pulse sequences, and so the TR can also be reduced.
2) The TR can also be reduced because flip angles less than 90º can be used (TR plays an important part in the time of the scan)

i.e. Gradient echo pulse sequences are associated with shorter scan times than spin echo pulse sequences.

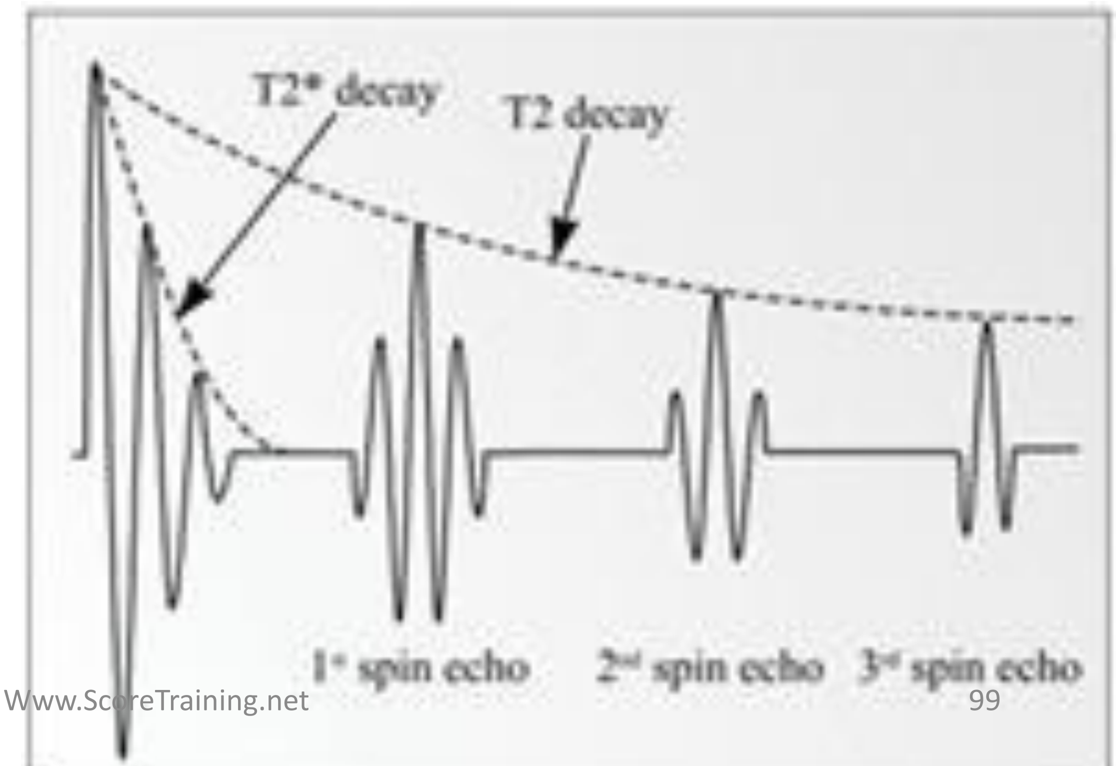


disadvantages of gradient echo pulse sequences:

No compensation for magnetic field inhomogeneities →Gradient echo pulse sequences are very susceptible to magnetic field inhomogeneities.

i.e. the T2* effects are not eliminated

In gradient echo pulse T2 weighting is termed T2* weighting



Weighting and contrast in gradient echo

TR and the flip angle control the amount of T1 relaxation that has occurred before the next RF pulse is applied.

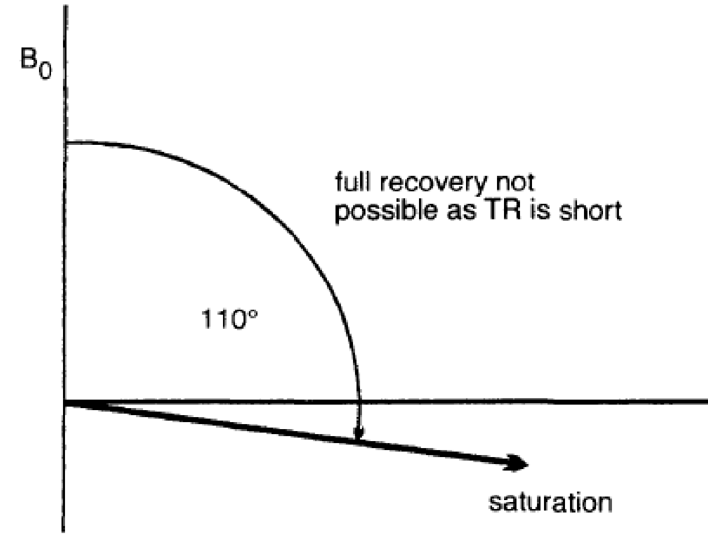
TE controls the amount of T2* decay that has occurred before the gradient echo is received by the coil.

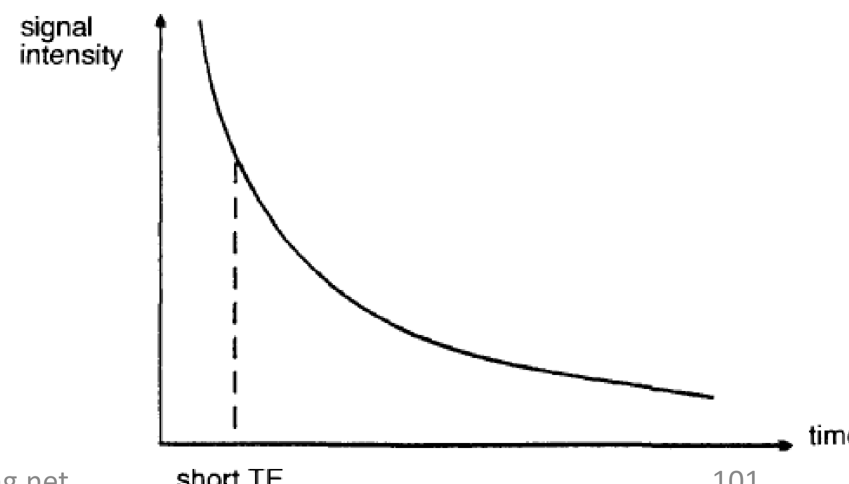
T1 weighting in gradient echo

- To obtain a T1 weighted image, neither the fat nor water vectors must have had time to recover full longitudinal magnetisation before the next RF pulse is applied (saturation) .

i.e. flip angle must be large and the TR short→fat and water vectors are still in the process of relaxing when the next RF is applied.

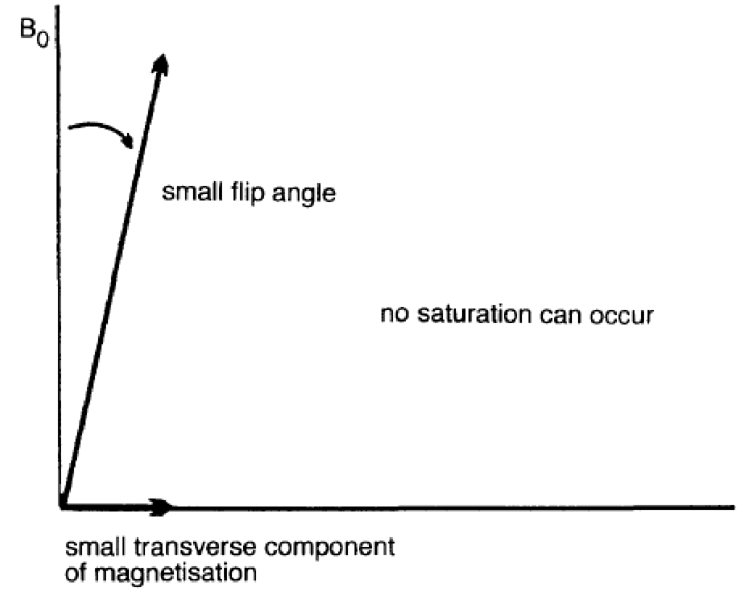
- To minimise T2* differences, the TE is short

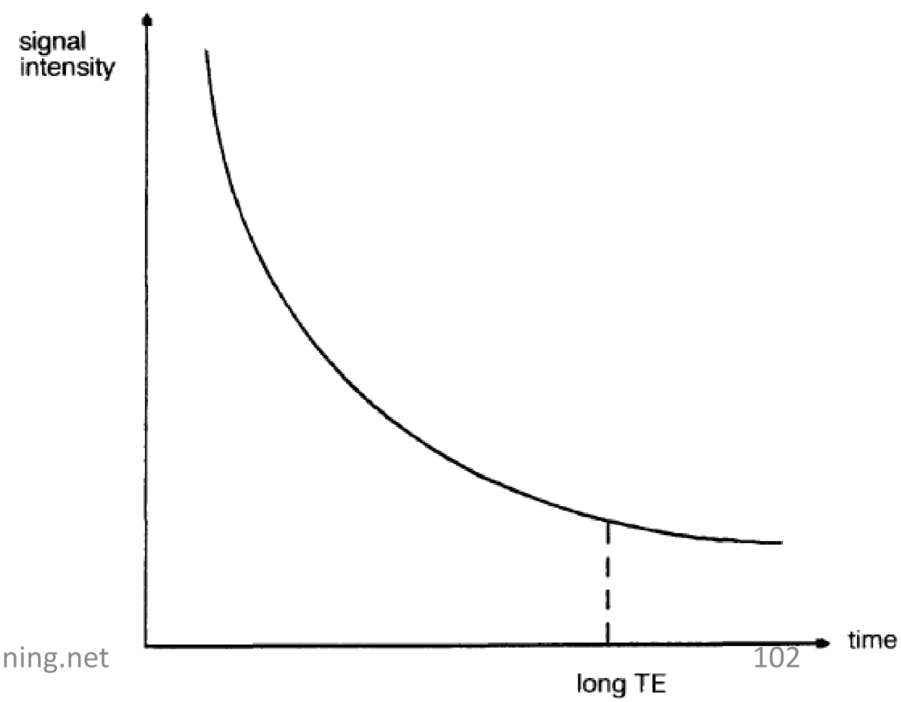




T2 * *weighting in gradient echo*

- To maximize T2* decay, the TE is long
- To minimize T1 recovery, the flip angle is small and the TR long





Proton *density weighting in gradient echo*

- To obtain a proton density weighted image both T1 and T2* processes are minimised
  - the TE is short
  - the flip angle is small
  - the TR is long

Typical values in gradient echo

Long TR	100 ms+
Short TR	less than 50 ms
Long TE	15-25 ms
Short TE	5-10 ms
Large flip angles	70°-110°
Low flip angles	5°-20°


Thank You!